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Biomechanics of knee osteoarthritis during stair negotiation: mechanisms and intervention strategies

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“***If your dreams do not scare you,***

they are not big enough”

Ellen Johnson Sirleef

Africa´s first female president

Contents

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B Dissemination of study findings

B.1 Submitted articles

Doslikova K, Maganaris CN, Baltzopoulos V, Luyten FP, Callaghan MJ, Jones RK, Felson DT, Reeves ND, and Verschueren SMP. Lower limb joint biomechanics during stair negotiation in patients with knee osteoarthritis compared to healthy controls. (under review with Journal of Biomechanics as of 04/07/2014, responses to reviewer´s comments submitted 16/02/2015)

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Callaghan MJ, Jones RK, Felson DT, and Reeves ND. The influence of a patellofemoral knee brace on knee joint kinetics and kinematics during stair negotiation in patients with knee osteoarthritis. (under review with Osteoarthritis and Cartilage as of 29/06/2014, responses to reviewer´s comments submitted 08/01/2015)

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Callaghan MJ, Jones RK, van Dieën JH, Felson DT, and Reeves ND. Is stair handrail use an effective strategy to reduce knee joint loads in osteoarthritis patients? (under review with Osteoarthritis and Cartilage as of 29/01/2015)

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Brown SJ, Callaghan MJ, Jones RK, Felson DT, and Reeves ND. Does knee osteoarthritis impair balance during stair negotiation? (under review with Journal of Biomechanics as of 07/03/2014)

B.2 Other associated submitted publications

Callaghan MJ, Guney H, Reeves ND, Bailey DI, Doslikova K, Maganaris CN, Hodgson R, and Felson DT. A knee brace alters patella position in patellofemoral osteoarthritis: A study using weight bearing magnetic resonance imaging. (under review with Osteoarthritis and Cartilage as of 04/03/2015)

B.3 Conference presentations

Doslikova K (2012): “Knee Osteoarthritis and Stair Negotiation – Year 1 Progress Update”. First Annual Move-Age Conference in Leuven, Belgium (oral presentation)

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Callaghan M, and Reeves ND (2013). “Knee Osteoarthritis and Stair Negotiation”. Student Research Conference at the MMU, Manchester, UK (poster presentation)

Doslikova K (2013): “Knee Osteoarthritis and Stair Negotiation – Year 2 Progress Update”. Second Annual Move-Age Conference in Leuven, Belgium (oral presentation)

Doslikova K (2014): “Knee Osteoarthritis and Stair Negotiation – Year 3 Progress Update”. Third Annual Move-Age Conference in Amsterdam, The Netherlands (oral presentation)

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Callaghan MJ, Jones RK, Felson DT, and Reeves ND (2014): “Knee Joint Biomechanics During Stair Descent in Patients With Knee Osteoarthritis vs. Controls”. World Congress on Osteoporosis, Osteoarthritis and Musculoskeletal Diseases in Seville, Spain (poster presentation)

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Callaghan MJ, Jones RK, Felson DT, and Reeves ND (2014): “The Influence of a Patellofemoral Knee Brace on Knee Joint Kinetics and Kinematics in Patients With Knee Osteoarthritis During Stair Negotiation”. World Congress on Osteoarthritis in Paris, France (poster presentation)

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Ireland A, Callaghan MJ, Jones RK, Felson DT, and Reeves ND (2014): “Knee Joint Kinetics and Kinematic Characteristics of Knee Osteoarthritis Patients During Stair Ascent”. World Congress of Biomechanics in Boston, USA (podium presentation)

B.4 Published abstracts

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Callaghan MJ, Jones RK, Felson DT, and Reeves ND. “Knee Joint Biomechanics During Stair Descent in Patients With Knee Osteoarthritis vs. Controls”. Proceedings of the 2nd World Congress on Osteoporosis, Osteoarthritis and Musculoskeletal Diseases, Seville, Spain. P. 404

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Callaghan MJ, Jones RK, Felson DT, and Reeves ND. “The influence of a patellofemoral knee brace on knee joint kinetics and kinematics in patients with knee osteoarthritis during stair negotiation”. Osteoarthritis and Cartilage, April 2014, Volume 22, Supplement, Page S89.

Doslikova K, Maganaris CN, Baltzopoulos V, Verschueren SMP, Luyten FP, Ireland A, Callaghan MJ, Jones RK, Felson DT, and Reeves ND (2014): "Knee Joint Kinetics and Kinematic Characteristics of Knee Osteoarthritis Patients During Stair Ascent". Proceedings of the 7th World Congress of Biomechanics, Boston, USA.

C Thesis abstract

In this thesis I evaluated a mixed knee OA group with medial tibiofemoral (TF) and patellofemoral (PF) OA and healthy age- and BMI-matched controls during stair negotiation. Participants were examined walking on a seven-step staircase with four embedded force platforms and kinematic data were recorded using a ten-camera motion capture system. Biomechanics of lower limb joints and balance control during this task was compared between groups. Secondly, strategies aimed at reducing the pain and task difficulty were evaluated, namely the use of a PF knee brace and use of the stair handrail. Results showed that the OA group negotiated stairs with a reduced knee joint moment in the sagittal plane, but a greater knee joint moment in the frontal plane compared to controls. No major balance differences were identified between the groups, with the exception of a greater centre of mass velocity in the OA group in the final phase of lowering during stair descent. A PF brace reduced the knee joint moment in the sagittal plane during stair ascent in OA patients, which would be expected to reduce PF loading. In OA patients, stair handrail use on the contralateral side to the affected knee reduced frontal plane knee joint moments, with implications for TF loading and handrail use regardless of side reduced sagittal plane knee joint moments, with implications for PF loading. This thesis adds to the current knowledge about knee OA and stair negotiation. This body of work has identified biomechanical alterations at the knee joint in a mixed knee OA population, which do not impact markedly upon balance control during stair negotiation. Intervention strategies involving PF bracing and use of stair handrails offer simple, but effective strategies for altering biomechanical demands on the knee and mitigating disease progression.

Chapter 1

Introduction

1.1 Background

Definition of OA, knee OA and disease epidemiology

Osteoarthritis (OA) is a chronic progressive degenerative disease and the most common form of arthritis affecting joints, especially load-bearing joints such as the knee. The prevalence of knee OA increases with age and since the global population is ageing, the management of the disease presents a socioeconomic burden[1-3]. Approximately 12 % of the American population over 60 years and 10 % of the British population over 55 years suffer from symptomatic knee OA[4-6]. However, there is a mismatch between symptomatic and radiographic knee OA as the clinical picture does not always fully correspond with the radiographic findings and vice versa[7,8]. For example, knee OA based upon radiographic findings was present in 19 % and 28 % of adults over 45 years in the Framingham and Johnston County studies, respectively, and in 37 % of adults over 60 years in the NHANES III study, which are higher prevalences compared to knee OA based upon symptoms in the same studies that was around 10 – 12 %[4,9,10]. Essentially, these studies indicate that many older people may have radiographic evidence of knee OA, without experiencing any symptoms.

Pathophysiology

The disease is characterised by an imbalance between the synthesis and degradation of the articular cartilage[11]. In knee OA, the process of cartilage remodelling is unbalanced and leads to pathological changes in the joint[12]. This process affects the cells of the articular cartilage, chondrocytes, which produce major components of the extracellular matrix in response to deterioration, collagen and proteoglycan. Changes in the chondrocytes are connected to abnormal proliferation and apoptosis in the cartilage[13]. Changes in the subchondral bone, the neighbouring layer to the cartilage, also play a role in cartilage degradation[14].

The disease affects the whole knee joint together with the surrounding structures. It leads to focal and progressive hyaline articular cartilage loss, changes in the subchondral bone, development of marginal outgrowth (osteophytes), increased thickness of the bony envelope (bone sclerosis), soft tissues changes in and around the joint, such as inflammatory changes in the synovium or tendency towards an increased laxity of ligaments, and muscular weakness[2].

Types of knee OA and symptoms

The knee is a tri-compartmental joint comprising the medial tibiofemoral (TF), lateral TF and patellofemoral (PF) joint. Knee OA can affect different compartments within the knee, with the medial TF and PF compartments being most frequently involved[15]. Much of the biomechanical research has focused on the development of TFOA and risk factors for disease progression reflecting the fact that the medial TF compartment is more commonly affected than the lateral compartment[4]. The predominant focus on TFOA also relates to the fact that there are specific biomechanical mechanisms that have been linked with medial TFOA progression as will be discussed in the following sections on **"Aetiology and risk factors"** and **"Knee OA, gait and stair negotiation"**. However, a growing body of literature has emerged about PFOA in the recent years to enhance understanding of this specific knee OA subgroup and stimulating a series of interventions and countermeasures to reduce pain. It was estimated that radiographic PFOA represents 65 % of all symptomatic knee OA[16]. From a clinical perspective the disease also affects multiple compartments at the same time resulting in a mixed disease of TF and PFOA.

The main symptom and complaint of knee OA patients is pain. The disease is a debilitating condition that leads to the loss of functional independence and ultimately to a decreased quality of life[13,15]. However, it seems that the pain and functional status in knee OA patients deteriorate and progress slowly[17,18]. It has been shown that even mild findings on radiography in PFOA are more likely than in TFOA to result in symptoms[19] such as pain, stiffness and functional limitations, which severely impact on activities of daily living (ADL) such as walking, stair-climbing and housekeeping[15,20,21].

Aetiology and risk factors

The aetiology of the disease is complex with roots both in biochemistry and biomechanics. There is a growing body of evidence for systemic factors and local biomechanical factors being involved in the disease initiation. The systemic factors include age, sex, ethnic characteristics, bone density, oestrogen replacement therapy (in post-menopausal women) and genetics all play a role in the susceptibility to knee OA, with knee OA being more prevalent in elderly compared to the young, females compared to males and African-Americans compared to white persons[2,18].

The local biomechanical factors responsible for knee OA include obesity, joint injury, joint deformity, joint laxity, sports participation and muscle weakness that can all be responsible for affecting a specific knee site and influencing the disease severity[2,8]. Knee malalignment has been found to be an independent risk factor for knee OA progression with varus alignment increasing the risk of medial TFOA and valgus alignment increasing the risk of lateral TFOA[22]. Also, the degree of malalignment is predictive of physical function decline[23]. The increased loading in the medial TF compartment characterised by a larger external knee adduction moment (EKAM) leads to TFOA progression[24,25]. The EKAM is developed in the stance phase of walking and acts around the knee joint with a tendency to rotate the tibia medially with respect to the femur in the frontal plane and is primarily caused by a medially acting ground reaction force (GRF)[2,24,25]. In the case of PFOA, it is known that patella malalignment leads to PFOA progression[26] and it is expected that higher loading in the PF joint might also lead to pain and progression.

Management

The disease represents a problem for both the patients and care-givers not only as it is a painful and debilitating condition, but also due to the fact that there is no definitive cure for it and the treatment for end-stage disease is joint replacement[15]. Felson et al. (2) actually state that OA is the most common reason for knee replacement.

As mentioned before, the disease can affect any of the knee joint compartments individually or in the form of a mixed disease, and having a unique function per compartment, tailored management with the affected compartment in mind seems appropriate. Such a treatment is favourable given also the heterogeneity of the disease with respect to its aetiology, history and clinical manifestation[15]. According to Felson et al. (2), targeting and modification of the risk factors is particularly important in the management of the disease in the weight-bearing joints, such as the knee joint, as it might provide opportunities for the prevention of disease-related pain and disability.

For example, focusing on TFOA a varus knee alignment is often observed, which is associated with a high EKAM. The EKAM has been identified as being involved in the initiation and progression of medial TFOA[2,24,25]. The EKAM is considered to be a surrogate measure of the compression forces in the medial knee compartment[25]. Therefore interventions such as gait retraining to improve the dynamic joint alignment should be considered[27]. Reeves and Bowling (25) recently reviewed a number of conservative biomechanical intervention strategies that might reduce the EKAM and possibly the disease progression. Such interventions and methods are cost-effective and much simpler alternatives compared to more complex and expensive therapeutic interventions[28]. Lateral wedge insoles are effective mostly for early stage knee OA patients, but can cause some discomfort when the inclination angle is particularly high. Their mechanism of work is by shifting the centre of pressure (CoP) and GRF closer to the knee joint centre, which then decreases the moment arm of the force about the knee in the frontal plane and the EKAM[25]. Thin-soled, flexible shoes and walking barefoot reduce the EKAM compared to thick-soled and inflexible shoes, possibly working via the mechanism of a more flexible movement and better force application by the foot to the ground[25,29]. In order to directly modify gait characteristics and reduce the EKAM, strategies such as a toe-out gait, lateral trunk lean and the use of a walking aid on the contralateral side to the symptomatic knee joint can be used[25]. A toe-out gait involves the foot being externally rotated during walking. Adopting a toe-out gait reduced the frontal plane lever arm and EKAM in the early stance phase. However, these changes were associated with significant increases in the sagittal plane lever arm and flexion moment. The peak knee adduction lever arm and moment were also reduced in the late stance phase without corresponding changes in the sagittal plane[30]. Therefore, the toe-out gait seems to be beneficial in reducing especially the second EKAM peak[25]. The lateral trunk lean (which may be naturally adopted by patients) works by leaning of the upper body over the stance limb during walking, which shifts the centre of mass (CoM) laterally and the CoM moves closer to the CoP of the stance limb. This moves the GRF towards the knee joint centre and by doing so reduces the moment arm of the GRF leading to a reduction of the EKAM[25,31]. The use of a walking aid on the contralateral side to the symptomatic knee will be discussed in the section **"Knee OA, stairs and handrail use"** below. Lastly, the EKAM can be modified and reduced by interventions acting directly upon the knee, such as a knee brace, or indirectly, such as muscle strengthening of the thigh and hip muscles[25]. Valgus knee braces can produce an external abduction moment to oppose the EKAM and unload the medial compartment of the knee joint, but can be impractical for daily use for many patients due to their bulk and not being aesthetically pleasing. Lateral hamstring muscles and hip abductor muscles can produce an internal knee abduction moment to oppose the EKAM and provide a variety of options for exercise and muscle strengthening intervention programmes[25]. The idea of targeting hip muscles rather than only knee musculature is supported also by Thorp et al. (32), possibly offering effective and biomechanically based therapeutic options for medial TFOA. Similarly, exercise intervention consisting of training the triceps surae and quadriceps femoris muscle groups have also been shown to reduce the EKAM during inclined gait and should therefore be considered as potentially promising for medial TFOA management[33].

There is no consensus on treatment of PFOA. Conservative, non-surgical interventions are the first line of treatment and are of paramount importance as surgical interventions are far less successful than in the case of TFOA. In addition, it should not be assumed that treatment plans for TFOA are applicable for PFOA[34,35]. For example, biomechanical management strategies that are effective for TFOA are not necessarily effective for PFOA, since different mechanisms are responsible for their effects. It has been shown that altered PF loading leads to symptoms and structural progression. Therefore, interventions to reduce the PF load seem appropriate to lessen symptoms and/or slow progression[34,35]. Such interventions include quadriceps strengthening, retraining of vastus medialis, glutei and trunk muscles, foot orthoses and bracing[35]. According to Hinman et al. (36), therapeutic knee taping is also effective in the management of pain and disability. In theory, these interventions could possibly enhance the alignment of the knee joint, redistributing forces through the knee joint and altering muscle co-activations, which may work against the disease symptoms and factors involved in disease progression. However, all of the above mentioned strategies to reduce the stress in PF compartment and limit the structural progression need numerous clinical trials before their effectiveness is established[35].

Felson et al. (28) also state that exercise plays an important role in primary, secondary and tertiary prevention. There are three different categories of exercise used in the management of knee OA: range of motion (ROM) and flexibility exercise, muscle conditioning and aerobic cardiovascular exercise. Such exercise interventions aim to improve cartilage nutrition and health, improve the loading capacity of the joint, increase functionality and reduce symptoms associated with the disease such as pain and disability[28]. Also, as previously mentioned, some studies have showed the beneficial effects of specific strengthening exercise programmes in terms of reducing the EKAM[25,32,33].

Finally, weight loss is an important and recommended therapeutic approach in patients with knee OA[37]. A study from 2005 showed that one pound of weight loss leads to a 4-fold reduction of knee load per step during ADL in overweight and obese older adults with knee OA[38]. According to Sharma et al. (23), obesity is indeed most strongly linked with OA at the knee joint[23]. The authors found that body mass index (BMI) was related to OA severity in patients with varus alignment, but not in those with a valgus alignment, also varus alignment explained much of the effect of the BMI on the disease severity. Therefore varus alignment seems to be one local biomechanical factor that may assist in deteriorating the knee most vulnerable to the effects of obesity. A combination of moderate exercise and weight loss is suggested as the cornerstone for the treatment of overweight patients with knee OA[39].

In order to reduce pain and disability in knee OA patients, the treatment options do not include exercise only, but range from nutriceuticals and chondrocyte transplantation, anti-inflammatory medication to health education[2]. An appropriate treatment for knee OA combines exercise with one or more oral agents and other biomechanical techniques. It is concluded that the goal of the knee OA management today remains the control of pain and improving functionality and health-related quality of life[28].

Knee OA, gait and stair negotiation

As mentioned previously, the medial TFOA is most commonly affected and so it has attracted a lot of attention in the recent years related to the initiation of the disease and mechanisms involved. During level walking, the EKAM, which has been described and discussed previously, is higher in medial TFOA patients compared to age-matched controls[19,40]. Typically the peak EKAM is reported and considered to be a valid measure for medial TF loading. Another measure of medial knee loading is the EKAM angular impulse, which measures the area under the EKAM-time curve throughout the duration of the stance phase[41,42]. Therefore the EKAM impulse seems to provide a fuller picture about medial knee loading than the EKAM and so it has been reported to be better at differentiating between disease severities[41].

Certain ADL such as stair negotiation are challenging for people with knee OA[43], which remains physically challenging even after total knee replacement[44]. Stair negotiation generates a higher EKAM both in healthy and OA populations[43,45] compared to level walking. The peak EKAM was higher during stair ascent compared to descent in healthy young individuals[45]. Riener et al. (46) found that stair negotiation also generates a three times greater internal knee extension moment compared to level walking in healthy subjects. Similarly, one study in healthy participants assessed the PF loading during a single stepping-up task and found that the external knee flexion moment was higher and the peak PF contact force was even 8 times higher compared to level walking[47]. Another study examined healthy old and young participants during ascending stairs and ramps. The elderly had a reduced external knee flexion moment compared to the young during both tasks and an increased EKAM, potentially increasing their risk of developing medial TFOA[48].

In people with PFOA stair climbing is actually considered the most common activity causing pain[37]. Only one study has looked at biomechanical changes in people with PFOA or a mixed disease during stair negotiation and they found that patients with either isolated PFOA or mixed PF and TFOA had reduced internal knee extension moments compared to controls both during stair ascent and descent[49]. However, only a few studies have investigated stair negotiation in OA as opposed to level walking[19,40] or a stepping task on a flight of steps rather than a staircase[49]; and these studies did not focus on PFOA or a mixed disease, but rather on TFOA[50,51]. Therefore, there is a gap in the current literature to investigate differences in knee joint loading in patients with predominantly PFOA or a mixed disease, and secondly to explore possible underlying mechanisms through ankle and hip biomechanical alterations during stair negotiation, which will be addressed in Chapter 2 of this thesis.

Knee OA, stairs and knee bracing

In the case of PFOA, as mentioned before, it has been shown that higher loading in the PF compartment leads to symptoms and structural progression[35]. Therefore interventions that reduce PF loading such as retraining of specific muscles and bracing[34,35] seem appropriate to minimize symptoms and/or slow disease progression. PF braces have the potential to reduce PF joint stress[15] and change patellar kinematics during static postures[52], which may give rise to reduced sagittal plane joint loading through a more optimal alignment of the patella during gait. There are no studies that have investigated the biomechanical effects of a PF brace as opposed to purely the symptomatic pain-relieving effects and no studies that have tested this type of brace on stairs, which is especially pertinent given that PFOA patients particluarly experinence pain and difficulties negotaiting stairs. Although a PF brace has recently been shown to reduce knee pain in persons with painful PFOA[34,53], it is unclear whether this results from any biomechanical effect on the knee joint. Therefore, there is a gap in literature about the effect of a PF brace on knee joint biomechanics and loading in a predominantly PFOA or a mixed knee OA population during stair negotiation, which will be addressed in Chapter 3 of this thesis.

Knee OA, stairs and handrail use

The use of walking sticks/canes during level walking has been tested as a potential strategy for unloading the painful joint. Research during level walking has shown benefits of contralateral walking stick use to the affected knee during level walking[54,55,56] by reducing medial knee loading, whereas the use of a walking stick on the same side as the affected knee actually increased medial knee loading[54]. The use of handrails during stair negotiation might have a similar impact on medial knee loading to that of using a walking stick on level ground. Additionally, it may also improve the ability for people with knee OA to negotiate stairs and enhance safety. For example, in older adults without knee OA, the use of handrails gave the participants additional perception of stability[57]. Light bilateral handrail use in healthy older adults has been shown to alter sagittal plane knee joint loading during stair ascent and descent compared to a condition without handrail use[58]. Currently, the effects of handrail use on knee loading in OA remains unknown, but is a very pertinent issue to investigate due to the markedly higher loads present during stair negotiation compared to level walking as mentioned above. If the previously reported findings of reduced sagittal plane moments in healthy older participants with handrail use[58] also hold true for a population with knee OA, this could be a beneficial strategy particularly for PFOA. Therefore, there is a gap in the literature for other novel strategies for effective management of the disease that could help reduce the joint moments and loading. The use of stair handrails could be a potentially easily applicable and inexpensive strategy to use in the knee OA population. The effects of handrail use on knee joint biomechanics during stair negotiation in individuals with mixed knee OA but predominant PF symptoms will be addressed in Chapter 4 of this thesis.

Knee OA, stairs and balance

The combination of pain and altered knee biomechanics in knee OA together with the high biomechanical demands during stair negotiation may lead to balance impairments in this population during this specific task. It is well established that measures of standing balance, for example by using a swaymeter, are poorer in knee OA patients compared to healthy controls[59]. Patients with moderate to severe knee OA seem to have more deficits in static and dynamic balance control than those with mild disease[60,61]. Furthermore, Duffell et al. (62) found that people with early medial TFOA had reduced postural stability on both their affected and unaffected limbs during single leg standing. During stair negotiation, patients with knee OA demonstrated less time in single support, greater step width and decreased total gait velocity compared with controls[63].

Gait stability can be assessed by investigating the movement and velocity of the CoM and CoP. The CoM-CoP separation indicates dynamic stability during gait[64,65]; the greater the separation, the greater the challenge to maintain balance[65]; a more lateral (or anterior) position of the CoM in relation to the CoP (i.e. greater CoM-CoP separation) leads to a more unstable body position in lateral (or anterior) direction. Similarly, a reduction in the CoM velocity might be a strategy of how to make gait more stable. There has been some research conducted using the CoM-CoP separation as a measure to quantify balance control in healthy populations[64,65]. A study investigating dynamic balance control in healthy elderly compared to young adults during stair descent showed that the elderly were at a greater risk of falls, possibly due to a reduced ability to safely control the motion of the CoM while moving down stairs by the knee extensors[66]. Also, handrail use has been shown to modify knee loading and improve aspects of safety during stair negotiation in healthy elderly[58]. Bilateral handrail use was found to alter the CoM-CoP separation compared to unaided stair descent in healthy elderly[58].

However, all of these aspects currently remain unknown in a population with knee OA. There are no data regarding potential balance impairment in a knee OA population on stairs, for example by investigating the above-mentioned CoM velocity or CoM-CoP separation. Secondly, it is unknown whether handrail use could improve balance control during stair negotiation in the OA population should balance impairment be present. These aspects will be examined in Chapter 5 of this thesis.

1.2 Thesis aim

The main aim of the PhD was to identify biomechanical factors contributing to the knee OA symptoms and progression related to the negotiation of stairs and biomechanical-based strategies to make stair negotiation easier for knee OA patients and to potentially slow disease progression. I believe that the work will significantly contribute to the current knowledge about knee OA and also add to the knowledge with regards to stair negotiation in this population as there are a series of gaps in the literature as highlighted above.

1.3 Thesis outline

The PhD has been divided into four studies to reflect upon the gaps in the literature identified above. There were four main research questions posed in this thesis:

1) Are there differences in lower limb biomechanics between a mixed knee OA population (PFOA and TFOA) compared to age- and BMI-matched healthy controls during stair negotiation (with exploration of possible underlying mechanisms explaining differences in knee joint biomechanics between the two groups, through ankle and hip biomechanical alterations)?

2) Does a PF knee brace alter knee joint kinetics and kinematics and potentially the knee joint loading during stair negotiation in patients with mixed knee OA?

3) Does stair handrail use alter knee joint biomechanics and potentially the knee joint loading during stair negotiation in the mixed knee OA population?

4) Is balance control compromised in mixed knee OA population compared to age- and BMI-matched healthy controls during stair negotiation and is the stair handrail use helpful in maintaining balance in the mixed knee OA population during stair negotiation?

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Chapter 2

Lower limb joint biomechanics during stair negotiation in patients with knee osteoarthritis compared to healthy controls

Under review with Journal of Biomechanics

2.1 Abstract

Objective: The aim of the study was to investigate biomechanical differences of the lower limb between knee osteoarthritis (OA) and control participants during stair negotiation.

Methods: Thirty male and female participants (58.9 ± 7.7 years) with mixed knee OA and predominantly PF symptoms and thirty healthy control participants were recruited. Participants ascended a 7-step staircase. Kinematic data were obtained by tracking the movement of rigid clusters and markers using a 10-camera motion analysis system (Vicon) and a modified 6 degrees of freedom full body model. Ground reaction forces (GRF) were measured from four force platforms embedded into steps. Joint moments were calculated through inverse dynamics by combining kinematic and GRF data. Pain was assessed using a visual analogue scale. Statistica software was used for the analysis.

Results: The key findings at the knee in the OA group compared to controls included a reduced peak internal extension moment during stair ascent by ~22 % and descent by ~17 % to reduce the load in the PF compartment; and an increased peak internal abduction moment during stair ascent by ~85 % reflecting the increased loading in the medial compartment. Stair descent was clearly more demanding for the OA group as they had more pain, a wider stride width and an increased peak internal abduction moment 1.7 times higher compared to ascent and therefore greater medial loading.

Conclusion: Mixed knee OA population with predominantly PF symptoms differed in lower limb biomechanical parameters when compared to healthy controls during stair negotiation.

2.2 Introduction

The knee is one of the most commonly affected sites by osteoarthritis (OA)[1]. Symptomatic knee OA is present in 12 % of Americans over 60 years and 10 % of British adults above 55 years[2-4]. The prevalence of radiographic knee OA in adults over 45 years was 19 % in Framingham and 28 % in Johnston County study, and 37 % among adults over 60 years in the NHANES III study[2,5,6].

Knee OA leads to pain, disability and a decreased quality of life[7,8]. The prevalence increases with age and since the population is ageing, the management represents a growing socio-economic burden[9]. Furthermore, there is no cure and the treatment for end-stage disease is joint replacement[7].

Different knee joint compartments can be affected such as the medial tibiofemoral (TF) and patellofemoral (PF) compartment[7]. The disease can be unicompartmental or a mixed disease. Much of the research focused on TFOA, but a growing body of work has emerged relating to PFOA. Radiographic evidence of PFOA corresponds well with symptoms[10], which is not true for TFOA[11], and even mild findings on radiography in PFOA impact activities of daily living (ADL)[7,12,13].

In the case of TFOA, certain biomechanical factors are involved in the progression of OA in the medial TF compartment such as the external knee adduction moment (EKAM)[9,14,15]. During level walking, the EKAM is higher in medial TFOA patients compared to controls[11,16].

Stair negotiation generates a higher EKAM both in healthy and OA populations compared to level walking[17,18]. The peak EKAM was higher during stair ascent compared to descent in healthy young individuals[18]. Aside from frontal plane differences between level walking and stairs, Riener et al. (19) found that stair negotiation generates an internal knee extension moment that is up to three times greater compared to level walking in healthy subjects. Similarly, one study in healthy participants assessed the PF loading during a single stepping-up task. They found that the external knee flexion moment was higher and the peak PF contact force was even 8 times higher compared to level walking[20]. In people with PFOA, stair climbing is considered the most common activity causing pain[21]. Only one study has looked at biomechanical changes in people with PFOA during stair negotiation[22]. Fok et al. (22) found that patients with either isolated PFOA or mixed PF and TFOA had reduced internal knee extension moments compared to controls both during stair ascent and descent. However, only few studies have investigated stair negotiation in OA as opposed to level walking[11,16] or a stepping task on a flight of steps rather than a staircase[22]; and these studies did not focus on PFOA[23,24].

The aim of this study was, therefore, to investigate differences in knee joint loading in patients with predominantly PFOA and explore possible underlying mechanisms through ankle and hip biomechanical alterations during stair negotiation. It was hypothesised that knee kinematics and kinetics will be altered in the OA compared to controls as a natural strategy to reduce the load in the painful PF compartment, namely that the OA group would negotiate stairs with less knee flexion and consequently a reduced internal knee extension moment. It was also hypothesised that there will be additional biomechanical changes in the hip and ankle to compensate for the knee since the lower limb can be seen as a linked kinetic chain and therefore adequate responses to altered knee kinetics and kinematics are expected to be reflected by altered hip and ankle biomechanics.

2.3 Methods

Recruitment

Thirty knee OA participants were recruited from local primary care centres. Thirty control (CTR) participants were recruited from the local region via retirement groups and Manchester Metropolitan University (MMU) staff. Ethical approval was obtained from the relevant bodies and written informed consent was obtained from all participants.

Inclusion criteria

Participants were included in the OA group if they were aged between 40 -70 years. All participants had mixed compartment OA with a Kellgren-Lawrence (K-L) score grade 2 or 3 in the PF joint which was greater than the K-L score for the TF joint of the same knee. The diagnosis was made in 21 participants by plain radiography. Nine participants had arthroscopic or MR imaging documented evidence of the mixed disease severity and distribution. Pain must have been present daily for previous 3 months and rated a score equal to or above 40 on a 0-100-mm Visual Analogue Scale (VAS) on the day of the assessment for a nominated aggravating activity. In 10 out of the 30 knee OA participants there was radiographic and symptomatic evidence of a bilateral disease. The CTR group was recruited to match the OA group with respect to age and BMI.

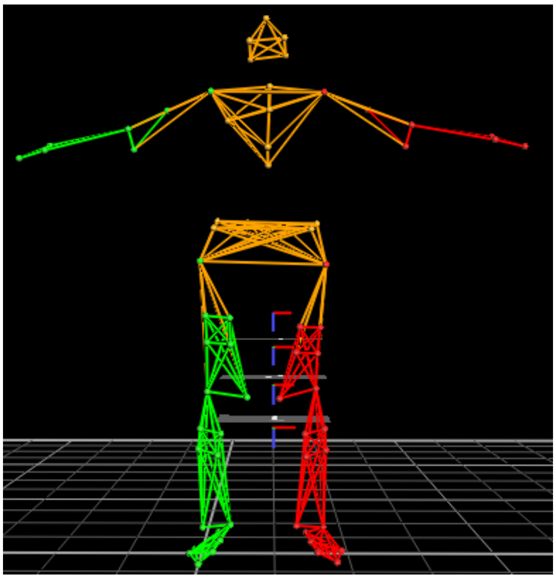
Exclusion criteria

Participants were excluded from the OA group if symptoms were traumatic in origin, if they had rheumatoid arthritis or other forms of inflammatory arthritis or if they had an intra-articular steroid injection into the painful knee in the previous month. Radiographs were read by a consultant musculoskeletal radiologist. Clinical assessments were made by an experienced physiotherapist. Participants were excluded from the CTR group if they had lower extremity problems or any other issues affecting their gait, such as recent injuries, neurological conditions, etc.

Experimental set up

A seven-step staircase instrumented with 4 individual force plates (Kistler, Winterthur, Switzerland; recorded at 1000 Hz, 300 x 500 mm), embedded into second, third, fourth and fifth steps, respectively, was used. Step dimensions represented standard dimensions with a going of 275 mm, riser height of 175 mm and width of 1050 mm. The handrails were at a height of 900 mm and in 31º of inclination. Participants wore a full body safety harness during the testing.

Kinetic and kinematic data were collected using force plates and a 10-camera VICON T20 optoelectronic motion analysis system (VICON motion systems Ltd., Oxford, UK), tracking a set of 69 passive retro-reflective markers using a modified 6 Degrees of freedom (6 DoF) whole body model developed by Capozzo et al. (25). The protocol was used to allow for segmental kinematics to be tracked in 6 DoF (Figure 1). This particular model enabled tracking of the lower limb segments irrespective of the position of the knee markers, which were removed after the static calibration was recorded. The knee joint was defined as the midpoint between the lateral and medial femoral epicondyle, but the movement of the knee joint and its rotational degreeds of freedom were tracked using the marker clusters attached onto the thigh and shin, reducing the problem of the movement of the soft tissue artefact (STA) at the knee when compared to the Plug In Gait model. Whilst STA is smaller on the thigh than the knee, there is still a degree of movement, especially as the muscles are continuously changing shape underneath and moving the clusters and/or the band holding them[26]. The hip joint was estimated using regression equations as part of the creation of the CODA pelvis[27]. The ankle joint was defined as the midpoint between the lateral and medial malleolus. This model allows for 6 DoF at the knee joint, hip and ankle joint, in 3 planes of rotation and 3 planes of translation.



Head markers:
- elastic head band with 4 markers

HEAD_BR	*..back right*
HEAD_FR	*..front right*
HEAD_BL	*..back left*
HEAD_FL	*..front left*

- 1 marker for static calibration only

HEAD_TOP	*..head top*

Trunk markers:

TOR_C7	*..7th cervical vertebra*
TOR_T10	*..10th thoracic vertebra*
TOR_CLAV	*..jugular notch*
TOR_STER	*..xiphoid process*
SCAPI	*..left scapula*
TOR_SHR/L	*..right/ left acromion*

Upper limb markers:

R/LARM_ELL	*..right/ left lateral elbow*
R/LARM_UP	*..right/ left middle upper arm*
R/LARM_ELM	*..right/ left medial elbow*
R/LARM_WRU	*..right/ left ulnar wrist*
R/LARM_WRR	*..right/ left radial wrist*
R/LARM_FIN	*..right/ left head of 2nd metacarpal bone*

Pelvic markers:

PEL_PSISL	*..left posterior superior iliac spine*
PEL_ASISL	*..left anterior superior iliac spine*
PEL_PSISR	*..right posterior superior iliac spine*
PEL_ASISR	*..right anterior superior iliac spine*
PEL_SAC	*..sacrum*
PEL_ILCL	*..left iliac crista*
PEL_ILCR	*..right iliac crista*
PEL_TROCHL	*…left greater trochanter*
PEL_TROCHR	*...right greater trochanter*

Lower limb markers:

R/LKNE_L	*..right/ left lateral femoral epicondyle*
R/LKNE_M	*..right/ left medial femoral epicondyle*
R/LTH_1, 2, 3, 4	*..right/ left 4 thigh markers on a cluster*
R/LANK_L	*..right/ left lateral malleolus*
R/LANK_M	*..right/ left medial malleolus*
R/LSHA_1, 2, 3, 4	*..right/ left 4 shin markers on a cluster*
R/LFOOT_H	*..right/ left heal*
R/LFOOT_1H	*..right/ left head of the 1st metatarsal bone*
R/LFOOT_5H	*..right/ left head of the 5th metatarsal bone*
R/LFOOT_1B	*..right/ left base of the 1st metatarsal bone*
R/LFOOT_5B	*..right/ left base of the 5th metatarsal bone*
R/LFOOT_TOE	*..right/ left tip of 2nd toe*

Figure 1. The static calibration image as captured in Vicon with the list of all the markers used. Markers were placed on specific anatomical landmarks either directly on the skin or onto tight fitting elastic clothing. There were 49 individual markers placed, 16 markers attached on 4 marker clusters (with 4 markers per cluster) for lower extremities and 4 markers on an elastic head band. Palpation was used to identify bony landmarks and the same researcher placed the markers in order to minimize the inter-researcher variability. The 4 lower extremity clusters were fastened by an elasticated band/ wrap.

Protocol

Participants were asked to ascend and descend stairs in a step-over-step manner at a speed controlled by a metronome at 90 beats per minute since kinetic variables can be influenced by differences in speed[28]. This speed was selected as it has previously been shown to correspond closely with the self-selected speed in elderly people during stair negotiation (92 + 10 steps/min)[29]. Three trials, i.e.: three ascents and descents, were recorded and stance phases were analyzed for the affected leg (or the more severely affected leg in case of bilateral disease) in the OA group and standardised to the right leg in the CTR group. The mean of six stance phases (two stance phases per trial; three trials) was used for the analysis. Participants were instructed not to hold onto the handrails unless needed and none of the patients used the handrails.

Data analysis

Kinematic and kinetic data recorded using the VICON system were labelled in VICON Nexus (Vicon Nexus 1.8.2). Post-processing calculation of kinematic and kinetic data was conducted using Visual3D software (Visual3D Student Edition v4.96.9; C-Motion Inc., Rockville, MD, USA). All lower extremity segments were modelled as rigid bodies. Anatomical frames were defined by landmarks positioned at medial and lateral borders of the joint, and for each segment a right handed co-ordinate systems was defined. Joint kinematics were calculated using an X–Y–Z Euler rotation sequence, where the X-axis was for the frontal, Y-axis for the sagittal and Z-axis for the transverse plane. Joint kinetic data were calculated using three-dimensional inverse dynamics, and exported internal joint moment data were normalised to body mass (Nm/kg). Pain was assessed using a 0-100-mm VAS after each condition was finished for ascent and descent separately.

Statistical differences between the groups (OA and CTR) for all parameters were compared separately for stair ascent and descent, and were tested using analysis of covariance (ANCOVA) with gait speed as a covariate. Walking speed was used as a covariate in case participants did not succeed in meeting the timing dictated by the metronome. Additionally, differences between stair ascent and descent for knee angles and moments were compared for each group using a paired samples Student's *t*-test.

2.4 Results

Participant characteristics

As presented in Table 1 both groups were similar with regard to age, height, weight and BMI. The OA group had the following evidence of mixed disease: The K-L grade 1/2/3 in the PF joint was 3.3/20/46.7 % of the sample respectively. The K-L grade 1/2/3 in the TF joint was 6.7/20/43.3 % of the sample respectively. The medial joint space narrowing (JSN) grade 1/2 was 30/40 % of the sample respectively. The lateral JSN grade 0/1/2 was 13.3/50/6.7 % of the sample respectively. The patellar JSN grade

1/2/3 was 43.3/20/6.7 % of the sample respectively. Information about the remaining 30 % of the sample was documented by arthroscopy or MR imaging.

Table 1. Participant Characteristics*.

	OA	CTR	*p*
	n= 30	n= 30	
Age, years	58.9 (7.7)	61.6 (11.7)	*0.290*
Height, m	1.7 (0.1)	1.7 (0.1)	*0.450*
Body mass, kg	77.0 (15.9)	74.0 (10.9)	*0.409*
BMI, kg/m^2	27.4 (3.8)	25.8 (3.0)	*0.094*
Sex	57 % F	47 % F	

* Data are presented as mean (SD) except for sex. The p-value corresponds to an independent t-test comparing the two groups.

ǂ Significant difference between groups ($p < 0.05$).

Stair ascent

Gait speed, stride width and pain

No differences were found between groups when comparing the gait speed and stride width during stair ascent. However, the OA group was in more pain compared to controls (Table 2).

Table 2. OA vs. CTR Group During Stair Ascent at Standardized Speed*.

		Mean (SD)		*p*
		OA	CTR	
		n = 30	n = 30	
Gait Speed (m/s)		0.50 (0.05)	0.51 (0.03)	*0.163*
Stride Width (m)		0.11 (0.02)	0.10 (0.02)	*0.078*
VAS (mm)		**24 (24)**	**0 (0)**	***0.000 †***
Hip	MIN Flexion Angle (^{0})	8.5 (10,1)	5.5 (10.0)	*0.344*
	MAX Flexion Angle (^{0})	64.6 (9.9)	60.3 (10.7)	*0.145*
	MAX Abduction Angle (^{0})	3.5 (3.8)	2.1 (3.4)	*0.080*

	MAX Adduction Angle (^{0})	11.3 (5.6)	11.2 (4.2)	*0.917*
	Total ROM in the Sagittal Plane (^{0})	56.0 (5.3)	54.8 (4.8)	*0.262*
	Total ROM in the Frontal Plane (^{0})	14.8 (5.5)	13.3 (4.0)	*0.153*
	Peak Extension Moment (Nm/kg)	**0.62 (0.18)**	**0.60 (0.17)**	***0.019 †***
	Peak Abduction Moment (Nm/kg)	0.86 (0.15)	0.94 (0.13)	*0.062*
Knee	MIN Flexion Angle (^{0})	12.9 (5.6)	15.5 (4.5)	*0.054*
	MAX Flexion Angle (^{0})	76.0 (5.3)	78.6 (5.4)	*0.100*
	MAX Abduction Angle (^{0})	**3.7 (5.2)**	**6.2 (4.0)**	***0.033 †***
	MAX Adduction Angle (^{0})	**7.0 (6.9)**	**4.0 (5.7)**	***0.043 †***
	Total ROM in the Sagittal Plane (^{0})	63.1 (6.3)	63.1 (5.4)	*0.857*
	Total ROM in the Frontal Plane (^{0})	10.7 (5.1)	10.2 (4.4)	*0.581*
	Peak Extension Moment (Nm/kg)	**1.05 (0.23)**	**1.35 (0.20)**	***0.000 †***
	Peak Abduction Moment (Nm/kg)	**0.24 (0.17)**	**0.13 (0.11)**	***0.004 †***
Ankle	**MAX Dorsiflexion Angle (^{0})**	**16.0 (3.7)**	**19.9 (3.7)**	***0.000 †***
	MAX Plantar Flexion Angle (^{0})	22.3 (6.7)	19.5 (6.4)	*0.067*
	MAX Eversion Angle (^{0})	**1.9 (4.0)**	**2.1 (3.8)**	***0.033 †***
	MAX Inversion Angle (^{0})	9.5 (4.1)	10.2 (5.0)	*0.606*
	Total ROM in the Sagittal Plane (^{0})	38.4 (7.6)	39.4 (6.3)	*0.772*
	Total ROM in the Frontal Plane (^{0})	**11.4 (2.9)**	**12.3 (4.4)**	***0.033 †***
	Peak Plantar Flexion Moment (Nm/kg)	1.21 (0.14)	1.22 (0.10)	*0.789*
	Peak Inversion Moment (Nm/kg)	0.22 (0.14)	0.26 (0.10)	*0.244*

* Data are presented as mean (SD). † Significant difference ($p < 0.05$).

Joint angles

During stair ascent, no differences were found between the groups when comparing hip angles. Compared to controls the OA group had a reduced maximal knee abduction angle by 2.5^0 and an increased maximal knee adduction angle by 3.0^0. The OA group had a reduced maximal ankle dorsiflexion angle by 3.9^0 compared to controls (Table 2).

Joint moments

During stair ascent, the OA group had an increased peak internal hip extension moment by ~3 % compared to controls (Table 2). Furthermore, the OA group had a reduced peak internal knee extension moment by ~22 % and an increased peak internal knee abduction moment by ~85 % compared to controls (Figure 2A and 2B, Table 2). No differences were found between the groups in ankle moments (Table 2).

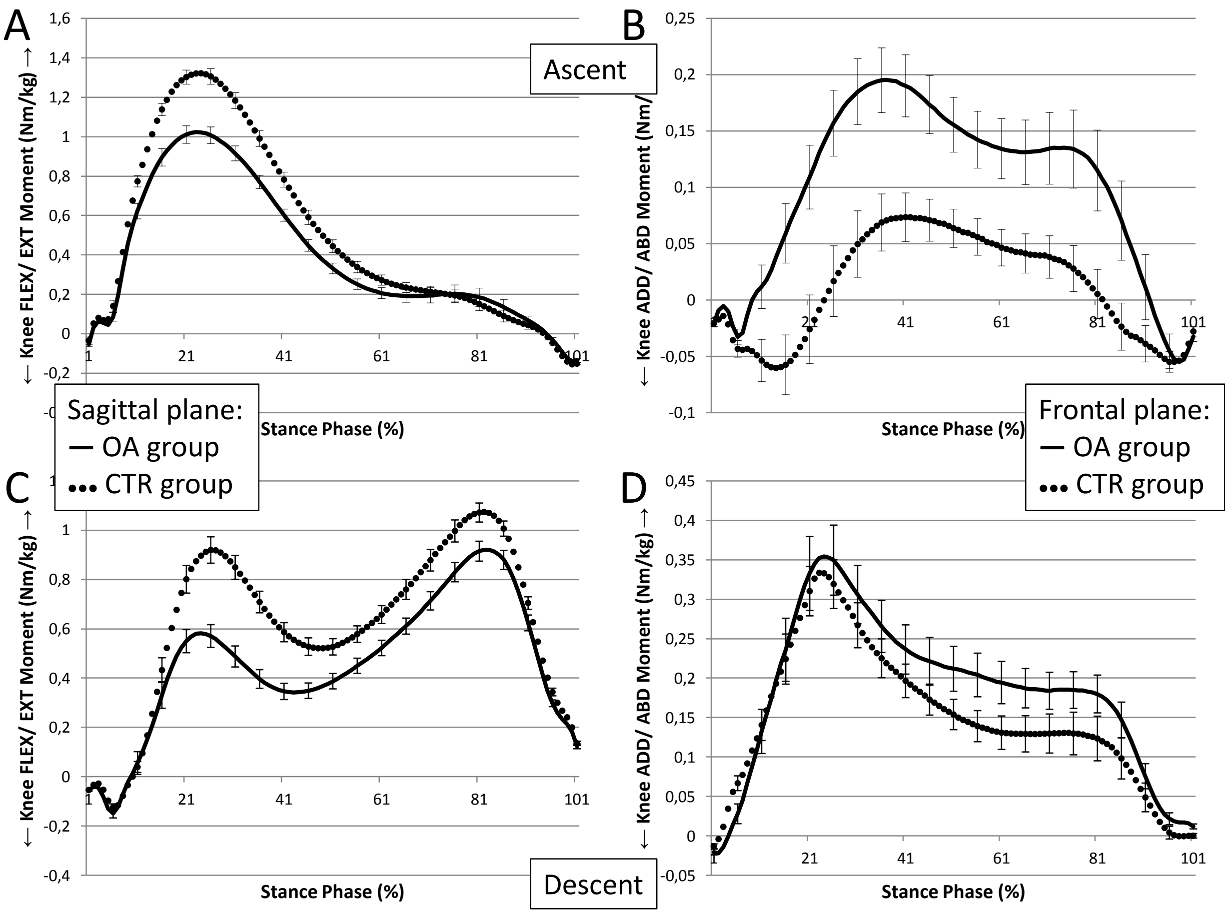


Figure 2. Mean internal knee joint moments for the OA (—) and CTR group (dotted) with standard error of the mean normalised to 100 % of stance phase at the standardized speed: sagittal knee joint moments during ascent (A), frontal knee joint moments during ascent (B), sagittal knee joint moments during descent (C), frontal knee joint moments during descent (D).

Stair descent

Gait speed, stride width and pain

Not only was the OA group in more pain compared to controls, but the OA group also descended stairs with a wider stride width and at a slower gait speed (Table 3).

Table 3. OA vs. CTR Group During Stair Descent at Standardized Speed*.

		Mean (SD)		*p*
		OA n = 30	CTR n = 30	
Gait Speed (m/s)		**0.49 (0.06)**	**0.52 (0.03)**	***0.009 †***
Stride Width (m)		**0.15 (0.02)**	**0.14 (0.03)**	***0.043 †***
VAS (mm)		**31 (29)**	**0 (0)**	***0.000 †***
Hip	MIN Flexion Angle (^{0})	13.4 (10.1)	13.2 (9.3)	*0.887*
	MAX Flexion Angle (^{0})	37.2 (9.7)	33.9 (9.3)	*0.175*
	MAX Abduction Angle (^{0})	**6.3 (4.3)**	**3.4 (3.3)**	***0.038 †***
	MAX Adduction Angle (^{0})	7.8 (4.3)	8.2 (4.1)	*0.919*
	Total ROM in the Sagittal Plane (^{0})	**23.8 (4.5)**	**20.6 (3.7)**	***0.007 †***
	Total ROM in the Frontal Plane (^{0})	**14.1 (3.9)**	**11.6 (4.3)**	***0.015 †***
	Peak Extension Moment (Nm/kg)	**0.51 (0.24)**	**0.56 (0.20)**	***0.011 †***
	Peak Abduction Moment (Nm/kg)	**1.06 (0.22)**	**1.24 (0.14)**	***0.003 †***
Knee	**MIN Flexion Angle (^{0})**	**13.0 (3.3)**	**16.4 (3.5)**	***0.001 †***
	MAX Flexion Angle (^{0})	96.5 (6.1)	98.1 (5.1)	*0.688*
	MAX Abduction Angle (^{0})	**3.0 (4.8)**	**5.1 (3.4)**	***0.026 †***
	MAX Adduction Angle (^{0})	**5.9 (5.8)**	**2.8 (5.2)**	***0.006 †***
	Total ROM in the Sagittal Plane (^{0})	83.6 (5.6)	81.7 (5.1)	*0.054*
	Total ROM in the Frontal Plane (^{0})	8.9 (4.6)	7.8 (3.6)	*0.184*
	Peak Extension Moment (Nm/kg)	**0.96 (0.23)**	**1.16 (0.19)**	***0.001 †***
	Peak Abduction Moment (Nm/kg)	0.41 (0.23)	0.36 (0.17)	*0.226*
Ankle	**MAX Dorsiflexion Angle (^{0})**	**23.6 (5.8)**	**28.6 (6.6)**	***0.012 †***
	MAX Plantar Flexion Angle (^{0})	24.9 (3.5)	23.9 (4.2)	*0.194*
	MAX Eversion Angle (^{0})	3.2 (3.8)	2.6 (3.1)	*0.527*
	MAX Inversion Angle (^{0})	8.8 (4.5)	8.6 (4.0)	*0.556*
	Total ROM in the Sagittal Plane (^{0})	48.5 (6.9)	52.5 (7.1)	*0.116*
	Total ROM in the Frontal Plane (^{0})	11.9 (2.7)	11.2 (2.3)	*0.053*
	Peak Plantar Flexion Moment (Nm/kg)	1.17 (0.14)	1.23 (0.14)	*0.072*
	Peak Inversion Moment (Nm/kg)	0.18 (0.11)	0.16 (0.10)	*0.161*

* Data are presented as mean (SD). † Significant difference ($p < 0.05$).

Joint angles

During stair descent compared to controls, the OA group had an increased total hip ROM in the sagittal plane by 3.2^0, an increased maximal hip abduction angle by 2.9^0 and an increased total hip ROM in the frontal plane by 2.5^0. The OA group had a reduced maximal knee flexion angle by 3.4^0, a reduced maximal knee abduction angle by 2.1^0 and an increased maximal knee adduction angle by 3.1^0 compared to controls. Also, the OA group had a reduced maximal ankle dorsiflexion angle by 5^0 compared to controls (Table 3).

Joint moments

During stair descent compared to controls, the OA group had a reduced peak internal hip extension moment by ~9 % and a reduced peak internal hip abduction moment by ~15 % (Table 3). Furthermore, the OA group had a reduced peak internal knee extension moment by ~17 % compared to controls. No differences between the groups were found in frontal plane knee moments or ankle moments (Figure 2C and 2D, Table 3).

Stair ascent compared to descent

In the OA group there was no difference between the gait speed during stair ascent and descent, however the stride width was greater and they had more pain during stair descent compared to ascent.

During stair descent in the OA group at the knee, there was an increased maximal flexion angle by 20.5^0, an increased total ROM in the sagittal plane by 20.5^0 and a reduced total ROM in the frontal plane by 1.8^0 compared to ascent (Table 4). See Table 4 for details for the CTR group.

Table 4. Stair Ascent Compared to Stair Descent*.

		Mean (SD)		*p*
		Ascent	Descent	
		n = 30		
OA	Gait Speed (m/s)	0.50 (0.05)	0.49 (0.06)	*0.406*
	Stride Width (m)	**0.11 (0.02)**	**0.15 (0.02)**	***0.000 †***
	VAS (mm)	**24 (24)**	**31 (29)**	***0.003 †***
	MIN Knee Flexion Angle (^{0})	12.9 (5.6)	13.3 (3.3)	*0.937*
	MAX Knee Flexion Angle (^{0})	**76.0 (5.3)**	**96.5 (6.1)**	***0.000 †***
	MAX Knee Abduction Angle (^{0})	3.7 (5.2)	3.0 (4.8)	*0.081*
	MAX Knee Adduction Angle (^{0})	7.0 (6.9)	5.9 (5.8)	*0.167*
	Total Knee ROM in the Sagittal Plane (^{0})	**63.1 (6.3)**	**83.6 (5.6)**	***0.000 †***
	Total Knee ROM in the Frontal Plane (^{0})	**10.7 (5.1)**	**8.9 (4.6)**	***0.017 †***
	Peak Knee Extension Moment (Nm/kg)	1.05 (0.23)	0.96 (0.23)	*0.097*
	Peak Knee Abduction Moment (Nm/kg)	**0.24 (0.17)**	**0.41 (0.23)**	***0.000 †***
CTR	**Gait Speed (m/s)**	**0.51 (0.03)**	**0.52 (0.03)**	***0.015 †***
	Stride Width (m)	**0.10 (0.02)**	**0.14 (0.03)**	***0.000 †***
	VAS (mm)	0 (0)	0 (0)	
	MIN Knee Flexion Angle (^{0})	15.5 (4.5)	16.4 (3.5)	*0.161*
	MAX Knee Flexion Angle (^{0})	**78.6 (5.4)**	**98.1 (5.1)**	***0.000 †***
	MAX Knee Abduction Angle (^{0})	**6.2 (4.0)**	**5.1 (3.4)**	***0.003 †***
	MAX Knee Adduction Angle (^{0})	**4.0 (5.7)**	**2.8 (5.2)**	***0.038 †***
	Total Knee ROM in the Sagittal Plane (^{0})	**63.1 (5.4)**	**81.7 (5.1)**	***0.000 †***
	Total Knee ROM in the Frontal Plane (^{0})	**10.2 (4.4)**	**7.8 (3.6)**	***0.000 †***
	Peak Knee Extension Moment (Nm/kg)	**1.35 (0.20)**	**1.16 (0.19)**	***0.000 †***
	Peak Knee Abduction Moment (Nm/kg)	**0.13 (0.11)**	**0.36 (0.17)**	***0.000 †***

* Data are presented as mean (SD). † Significant difference ($p < 0.05$).

In the OA group, the peak internal knee abduction moment was ~71 % greater during stair descent compared to ascent (Figure 3, Table 4), while it was ~177 % greater in the CTR group (Figure 3, Table 4). See Table 4 for more details.

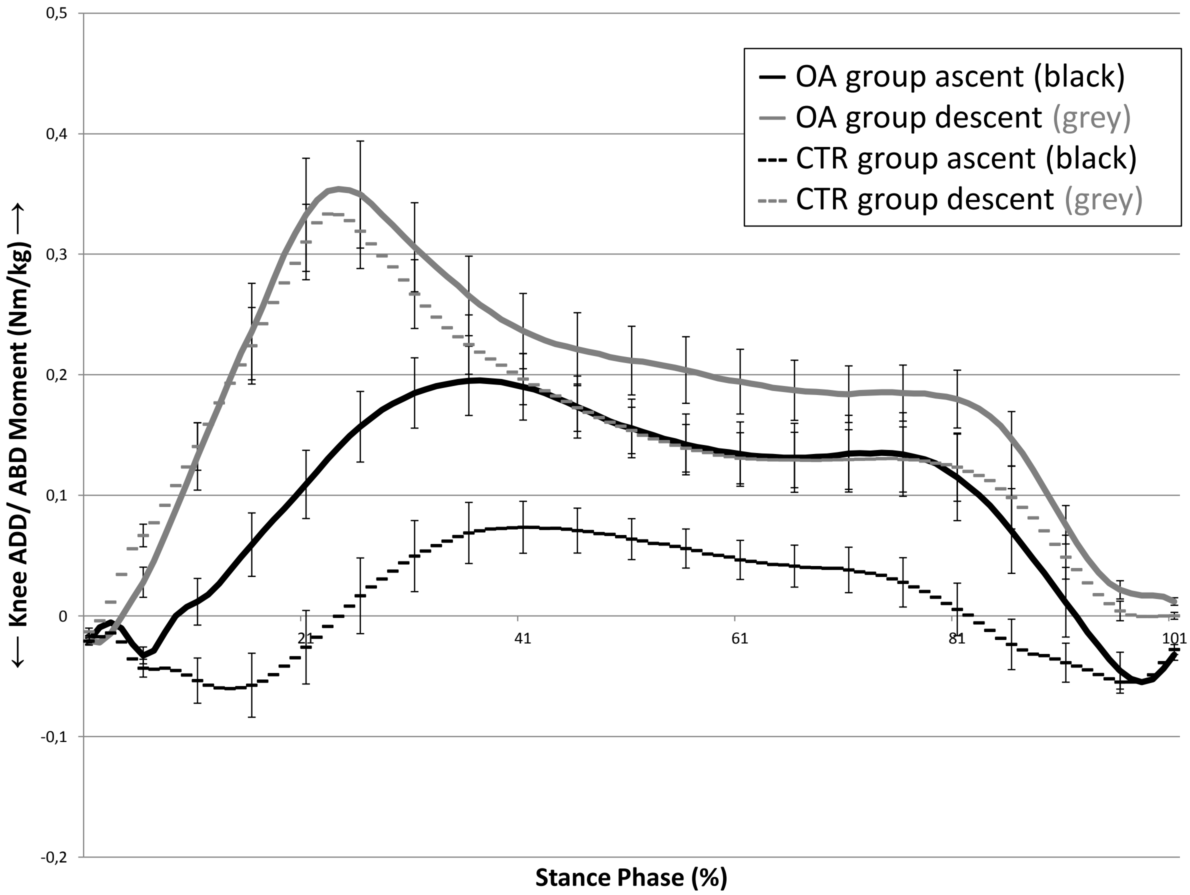


Figure 3. Mean frontal internal knee joint moments for stair ascent compared to stair descent with standard error of the mean normalised to 100 % of stance phase at the standardized speed: OA group ascent (—), OA group descent (—), CTR group ascent (---) and CTR group descent (---).

2.5 Discussion

This is the first study that investigated biomechanical characteristics in mixed knee OA participants with predominantly PF symptoms compared to healthy controls during stair negotiation on a seven-step staircase. The difference in knee pain during the task between the OA and controls was expected, however, the importance of the study lies in highlighting differences between the two groups at all three lower limb joints both during stair ascent and descent, as well as differences between ascent and descent in both groups.

Stair ascent

The key findings during stair ascent at the knee in the OA group compared to controls were a reduced peak internal knee extension moment confirming the hypothesis, which might have been a response to the pain in the PF joint, and secondly the increased peak internal knee abduction moment, which is in line with the literature of increased EKAM in the TFOA population during gait[11,16] since an increased external joint moment is matched similarly by an increased internal joint moment. Although I studied a mixed knee OA population, I observed similarly increased loading of the medial TF compartment as typically seen in the literature for medial TFOA. The peak internal knee abduction moment was increased by ~85 % in OA patients, which is a greater difference than observed between OA and control groups during level walking where there was an increase of the external knee adduction moment by about 30 %[11]. Therefore the results suggest that during stair ascent OA participants were able to reduce the PF load by decreasing the internal knee extension moment, but that was possibly done at the expense of an increased load on the medial knee compartment compared to controls. Similarly to the present study, lower internal knee extension moments in OA participants compared to controls were found both during level walking and stair ascent on a flight of four steps by Kaufman et al. (24) and it was suggested that it was their strategy to minimize pain by reducing the internal knee extension moment and so minimizing their PF loading.

The key difference in the OA group compared to controls at the hip was an increased peak internal hip extension moment during stair ascent, possibly as a result of a reduced peak internal extension moment at the knee. As the knee moment was reduced, the ground reaction force (GRF) vector was passing closer to the knee joint and simultaneously further to the hip joint, and as a consequence producing a greater hip moment. The key finding at the ankle was a reduced maximal dorsiflexion angle, which might be reflecting the way the OA group negotiated stairs: generally “stiffer” compared to controls.

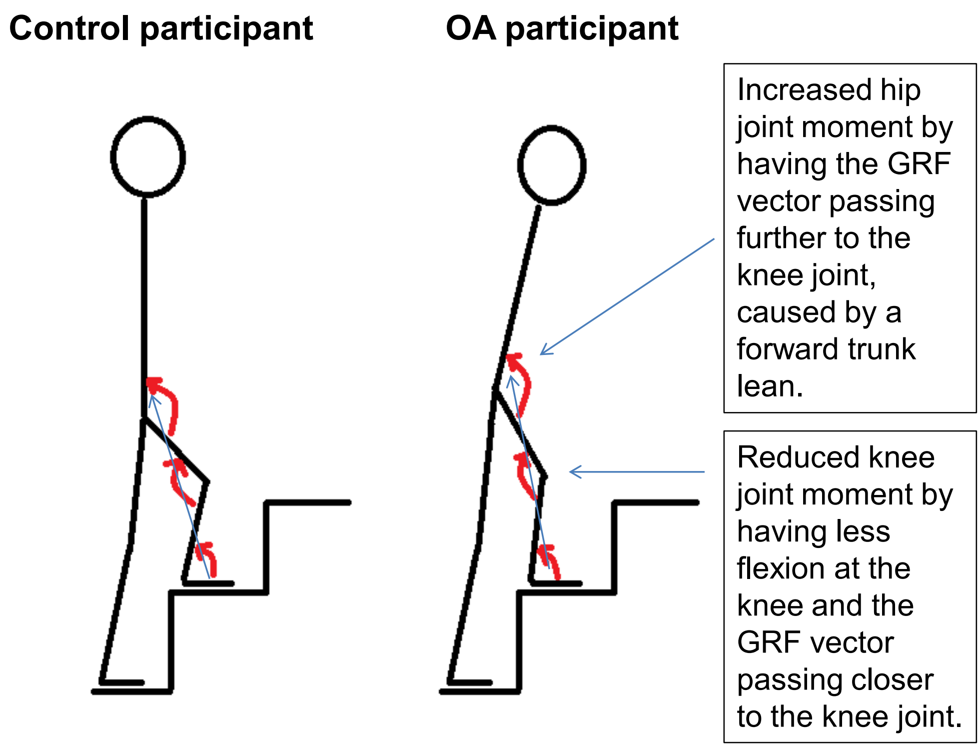


Figure 4. Schematic picture of stair ascent and the differences observed between the OA and control participants.

Stair descent

The key findings during stair descent at the knee in the OA group compared to controls was a reduced peak internal knee extension moment confirming the hypothesis, which again might be reflecting strategies to reduce the pain and load in the PF compartment in agreement with the findings by Kaufman et al. (24). Additionally, it may have been a way to put less demand on quadriceps muscle, which was shown to be weaker in both early and established knee OA[11]. Hinman et al. (30) found that subjects with OA had less knee flexion during early stance as in my study, and a delayed onset of vastus lateralis during stair descent (not during ascent) compared to controls. They concluded that these findings might effect force production during stair descent[30]. Similarly to the present study, their OA group consisted of patients who all had both PF and TFOA.

Although the frontal plane kinematics were different between the groups, there was no difference in the frontal knee moments. As stair descent is a more challenging task for knee OA population[17], this finding was not expected. However, the OA group descended stairs at a slower speed, which might have contributed to this finding.

The key differences in the OA group compared to controls at the hip was an increased total ROM in the sagittal and frontal plane, and a reduced peak internal hip extension and abduction moment; and at the ankle there was a reduced maximal dorsiflexion angle, which are again all together possible compensatory mechanisms to reduce the pain and movement at the knee.

The present study is novel with regard to knee OA and stair negotiation as I used a staircase with 7 steps as much simulating natural stair negotiation unlike previous studies. A recent study compared patients with PFOA, mixed disease of PF and TFOA and a CTR group during stair ambulation on a flight of 3 steps[22]. Compared to controls both OA groups had lower internal knee extension moments during stair ascent and descent, and during stair descent the OA groups had reduced knee flexion angles corresponding to my results. Baliunas et al. (16) compared medial TFOA group with controls during level walking. Apart from an increased peak EKAM, which is well established in the literature[11], they did not identify differences in peak sagittal plane knee moments between the groups[16] whereas in contrast, I found differences between groups in the sagittal plane moments. This probably reflects the task studied as compared to level walking and secondly the OA population in the present study having a mixed disease with predominantly PFOA symptoms and therefore showing differences in the sagittal plane. Kaufman et al. (24) compared OA group with controls during level walking and on a flight of four steps and apart from the differences in knee moments discussed above, they also found that knee flexion angles were greater on stairs compared to level walking. Another study looked at knee kinematics during stair negotiation on a four-step staircase and found a smaller maximal knee flexion in the OA group compared to controls during stair ascent possibly due to pain and stiffness as a compensatory mechanism to minimize quadriceps loading and thereby reduce compressive forces at the knee[23].

Gait speed

Although stair negotiation is a constrained task due to stair dimensions[31], I tried to standardize the speed. However, the OA group descended the stairs more slowly compared to controls. To confirm that I did not cause any alteration to the natural gait strategy by constraining speed, participants also walked at their spontaneously selected speed (data not shown). This analysis yielded similar results to findings presented.

Stride width

The OA group had a wider stride width compared to controls during stair descent. I hypothesise that this might be due to the fact that stair descent is more challenging and demanding in terms of medio-lateral balance. The wider stride resulted in greater knee moments in the frontal plane during descent compared to ascent as observed in the present study and others[17].

Stair ascent compared to stair descent

The key finding in the OA group was the peak internal knee abduction moment 1.7 times higher during stair descent compared to ascent, this difference was even greater in the CTR group with the moment being 2.8 times higher. This reflects the increased loading in the medial compartment during stair descent compared to ascent in both groups, as suggested by Guo et al. (17). In contrast, Hall et al. (18) identified that peak EKAM was higher during stair ascent compared to descent. However, in their study they only had a 3-step staircase and 16 healthy young individuals in comparison to my study[18]. Kaufman et al. (24) showed that internal knee extension moments were greater during stair descent compared to ascent. I failed to confirm this possibly reflecting the differences of the study groups and the set up.

Limitations and recommendations

There are possible limitations. Firstly, I had a mixed knee OA population, which makes interpretation of findings more complex since medial TFOA and PFOA are two different conditions[7]. However, mixed disease is often seen in clinics and therefore my study reflects clinical reality. Secondly, I did not examine or exclude participants with varus or valgus alignment, and malalignment is important for joint loading[32,33].

Thirdly, the model I used was possibly not sensitive enough to measure the movement of the knee joint in the frontal plane with high accuracy, especially considering the role of soft tissue artefact. Therefore the results, some of which are small in magnitude, need to be interpreted with caution. Additionally, the model used in the present study has limitations on its own. It does not use a functional axis of rotation determined during isolated motion capture trials at the knee joint, which is what some researchers have used[34] and which may describe what happens in the joint more closely compared to the present method or the geometry-based centre of rotation used in the Plug In Gait model. The present model allows for 6 DoF to be tracked at the knee, both the rotation and translation along three axes, which might have influenced the present results. The 6DoF model still has a STA due to the movements of the plates with respect to the underlying bones but the movement of anatomical markers on the knee will not be reflected in the knee kinematics as it is the case in the Plug In Gait model, which relies on the knee markers position for both the joint definition and tracking. In the 6DoF model the movement is tracked using clusters that are placed away from the joints in order to minimise skin movement over the bony prominences. The limitations of using a 6DoF model and specifically what the potential negative impact of translation at the knee joint might be upon the present results remains hypothetical. It probably means that our results were an overestimation of what actually happens at the knee joint, but this is only our speculation. The translation at the knee joint can affect the moment as it should account for the “sliding” of the knee axis, given it is not a fixed pivot system. This should be an advantage over the Plug In Gait model as it resembles the reality more closely. However, it adds extra complications to the model and possibilities for error.

Furthermore, by looking at the trunk kinematics and kinetics, I could have gained more insight into other compensatory mechanisms that the OA participants might have used as partly insinuated by the changes in the hip joint moments found.

Lastly, my OA group could be regarded as a relatively well-functioning OA cohort since participants were able to negotiate the stairs without handrail use, and further their body mass was not significantly higher than controls. Therefore, the study should be replicated in more severely affected OA groups and in larger studies in order to be representative of the wider knee OA population. Also, a global measure of function, for example the Western Ontario and McMaster Universities Arthritis Index, to compare my OA group to other knee OA study populations would have provided more insight into the patients´ functional status rather than only reporting structural findings, which do not necessarily reflect the full clinical picture[35].

The importance of the present study lies in underlining the fact that knee OA is a broad diagnosis that should be specified to a compartment and more studies should focus on either isolated PFOA or a mixed PFOA and TFOA as these patients represent different clinical realities from already quite extensively documented TFOA. The implication for patients with different compartmental involvement is to try to find new mechanisms of reducing increased joint loading when present and to explain the decreased loading when present, and possibly apply these mechanisms in disease management in a subject- and/or compartment-specific way.

With respect to future studies, pain-relieving effects of taping and bracing should be further examined during stair negotiation[7]. Secondly, the idea to reduce the load by insoles and braces should be further examined since there is a large variety of such devices. Lastly, there are compensatory mechanisms and gait retraining techniques that can reduce the load at the knee[15], such as forward trunk lean[36].

Conclusion

In conclusion, it was found that a mixed knee OA population with predominantly PF symptoms differs in lower limb biomechanical parameters when compared to healthy controls during stair negotiation.

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Chapter 3

The influence of a patellofemoral knee brace on knee joint kinetics and kinematics during stair negotiation in patients with knee osteoarthritis

Under review with Osteoarthritis and Cartilage

3.1 Abstract

Objective: The aim of this study was to investigate effects of a patellofemoral (PF) brace on knee biomechanics during stair negotiation in patients with knee osteoarthritis (OA).

Methods: Thirty participants with predominantly PFOA (40-70 years) negotiated a 7-step staircase wearing the brace (Ossur Bioskin Q brace) and in a control (CTR)

condition without the brace. Kinematic data were obtained by using a 10-camera motion analysis system (Vicon). Ground reaction forces (GRF) were measured from force platforms. Joint moments were calculated through inverse dynamics by combining kinematic and GRF data. A paired t-test was used to test for differences between conditions. Values are means followed by 95 % confidence intervals (CI).

Results: During stair ascent at the knee, the brace (BR) reduced the maximal flexion angle [BR: 73.3^{0}, CI 71.8-74.7^{0}; CTR: 76.0^{0}, CI 74.0-78.0^{0}], total sagittal plane range of motion (ROM) [BR: 61.1^{0}, CI 59.0-63.3^{0}; CTR: 63.1^{0}, CI 60.7-65.5^{0}], maximal adduction angle [BR: 4.9^{0}, CI 2.8-7.1^{0}; CTR: 7.0^{0}, CI 4.4-9.6^{0}], total frontal plane ROM [BR: 9.0^{0}, CI 7.0-10.9^{0}; CTR: 10.7^{0}, CI 8.8-12.6^{0}] and internal peak extension moment [BR: 1.00 Nm/kg, CI 0.92-1.09 Nm/kg; CTR: 1.05 Nm/kg, CI 0.96-1.13 Nm/kg] compared to the CTR condition. During stair descent at the knee, the brace reduced the maximal flexion angle [BR: 94.7^{0}, CI 92.7-96.8^{0}; CTR: 96.5^{0}, CI 94.3-98.8^{0}] and total sagittal plane ROM [BR: 82.1^{0}, CI 80.1-84.0^{0}; CTR: 83.6^{0}, CI 81.5-85.7^{0}] compared to the CTR condition.

Conclusion: PF knee bracing minimally yet significantly alters knee biomechanics in PFOA patients during stair negotiation.

3.2 Introduction

Osteoarthritis (OA) is a chronic degenerative disease affecting joints and it is the most common form of arthritis[1]. The prevalence of the disease is high and increases with age. Furthermore, as the global population is ageing, the treatment and management of OA represents a growing health and socioeconomic burden[1]. There are 12.1 % of Americans over 60 years with symptomatic knee OA[2] and painful knee OA is reportedly present in up to 10 % of adults aged above 55 years in the UK[3]. As there is no definitive cure for the disease and the treatment for end-stage disease is joint replacement[4], there is a need for thorough investigation and utilization of effective conservative treatments.

There is a growing body of literature focusing on the sub-group of knee OA which affects the patellofemoral (PF) joint. This important sub-group has often been forgotten despite the PF joint being more likely than the tibiofemoral (TF) joint to result in knee OA symptoms such as pain and stiffness as well as functional limitations[5], which severly impact on activities of daily living (ADL)[4,6]. Certain ADL such as stair negotiation are challenging for people with knee OA[7]. Additionally, stair negotiation is considered the most common activity causing pain in people with PFOA[8]. Stair negotiation generates higher frontal plane knee joint moments compared to level walking[7] and the internal knee extension moment is up to three times greater compared to that during level walking[9]. A recent study found that patients with either isolated PFOA or mixed PF and TFOA had reduced internal knee extension moments compared to a control group both during stair ascent and descent[10].

There is no consensus on treatment of PFOA. Conservative, non-surgical interventions are the first line treatments and are of paramount importance, as surgical interventions are far less successful for PFOA than in the case of TFOA. In addition, it should not be assumed that treatment plans for TFOA are applicable to PFOA[11,12]. It has been shown that higher loading in the PF compartment leads to symptoms and structural progression[12]. Therefore interventions that reduce PF loading such as retraining of specific muscles and bracing[11,12] seem appropriate to minimize symptoms and/or slow disease progression. PF braces have the potential to reduce PF joint stress[4] and change patellar kinematics during static postures[13], which may give rise to reduced sagittal plane joint loading through a more optimal alignment of the patella during gait. Additionally, therapeutic knee taping has been shown to be effective in the management of pain and disability in knee OA patients[14]. The mechanisms through which PF loading and pain may be reduced with bracing (and taping) are unclear, but restrictions to sagittal plane knee joint range of motion due to a greater perception of joint stability, with subsequent reductions in sagittal plane joint moments, may play a role.

The aim of this study was, therefore, to investigate the effects of a PF brace on knee joint biomechanics during stair negotiation in individuals with predominant PFOA. There are no studies that have investigated the biomechanical effects of a PF brace as opposed to purely the symptomatic pain-relieving effects and no studies that have tested this type of brace on stairs. Although this brace has recently been shown to reduce knee pain in persons with painful PFOA[11,15], it is unclear whether this results from any biomechanical effect on the knee joint. It was hypothesised that the PF brace would alter sagittal plane knee kinematics and partly as a consequence reduce sagittal plane joint moments about the knee, such as the internal knee extension moment, during stair negotiation.

3.3 Methods

Participants

Thirty participants were recruited from local primary care centres, hospital-based orthopaedic, rheumatology and physiotherapy clinics. All participants had participated in a larger randomised clinical trial that involved twelve weeks wearing a PF brace; the present study constitutes a separate biomechanical study of this cohort approximately 6 months after they had completed the trial; participants were therefore not brace-naïve. All of the participants continued to use the brace intermittently as their symptoms demanded. Participants had predominantly PFOA. The ethical approval was obtained from the Central Manchester Local Research Ethics Committee (LREC) and the Manchester Metropolitan University. Written informed consent was obtained prior to the testing from all the participants.

Inclusion criteria

Participants were included if they were aged between 40 -70 years, had a Kellgren-Lawrence (K-L) score grade 2 or 3 in the PF joint greater than the K-L score for the TF joint of the same knee. Participants needed to have PF joint symptoms such as pain reproduced with stair climbing, kneeling, prolonged sitting or squatting and lateral or medial patellar facet tenderness on palpation or a positive patellar compression test. This clinical test involved the patient contracting their quadriceps fully whilst their patella was compressed against the femur. Pain must have been present daily for the previous 3 months and rated a score equal to or above 40 on a 0-100-mm Visual Analogue Scale (VAS), with 0 indicating no pain and 100 indicating the maximum pain imaginable, on the day of the assessment for a nominated aggravating activity, which usually involved stairs or prolonged sitting. In 10 out of the 30 knee OA participants there was radiographic and symptomatic evidence of a bilateral disease, with the side that was examined for the effects of brace use in the present study being the most severely affected.

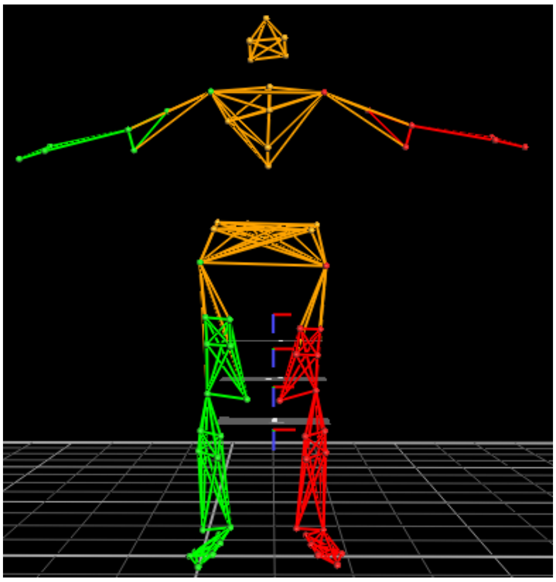
Exclusion criteria

Participants were excluded by an experienced musculoskeletal physiotherapist if the predominant symptoms emanated clinically from the TF joint, from meniscal or ligament injury, if they had rheumatoid arthritis or other forms of inflammatory arthritis or if they had an intra-articular steroid injection into the painful knee in the previous month based on the medical records provided. All radiographs were read by one consultant musculoskeletal radiologist. If the same scans were rated by two or more radiologists there is the possibility that classifications might have been slightly different, but this would not have affected the main outcome of the study. Clinical assessments were made by an experienced musculoskeletal physiotherapist.

Experimental set up

A seven-step staircase instrumented with 4 individual force plates (Kistler, Winterthur, Switzerland; recorded at 1000 Hz, 300 x 500 mm), embedded into the second, third, fourth and fifth steps respectively, was used in the study. The step dimensions represented standard stair dimensions with a going of 275 mm, a riser height of 175 mm and a width of 1050 mm. The handrails were at a height of 900 mm above the step and in 31° of inclination on both sides. All participants wore a full body safety harness attached to a safety beam on the ceiling and safety line on the floor connected by rope secured by the harness operator during the stair testing as a safety precaution.

Kinetic and kinematic data were collected using the force plates and a 10-camera VICON T20 optoelectronic motion analysis system (VICON motion systems Ltd., Oxford, UK), tracking a set of 69 passive retro-reflective markers using a modified 6 Degrees of freedom (6 DoF) whole body model developed by Capozzo[16]. This kinematic model ensured that no markers needed to be placed on the brace during the stair trials and allowed for segmental kinematics to be tracked in 6 DoF (Figure 1).



Head markers:
- elastic head band with 4 markers

HEAD_BR	*..back right*
HEAD_FR	*..front right*
HEAD_BL	*..back left*
HEAD_FL	*..front left*

- 1 marker for static calibration only

HEAD_TOP	*..head top*

Trunk markers:

TOR_C7	*..7th cervical vertebra*
TOR_T10	*..10th thoracic vertebra*
TOR_CLAV	*..jugular notch*
TOR_STER	*..xiphoid process*
SCAPI	*..left scapula*
TOR_SHR/L	*..right/ left acromion*

Upper limb markers:

R/LARM_ELL	*..right/ left lateral elbow*
R/LARM_UP	*..right/ left middle upper arm*
R/LARM_ELM	*..right/ left medial elbow*
R/LARM_WRU	*..right/ left ulnar wrist*
R/LARM_WRR	*..right/ left radial wrist*
R/LARM_FIN	*..right/ left head of 2nd metacarpal bone*

Pelvic markers:

PEL_PSISL	*..left posterior superior iliac spine*
PEL_ASISL	*..left anterior superior iliac spine*
PEL_PSISR	*..right posterior superior iliac spine*
PEL_ASISR	*..right anterior superior iliac spine*
PEL_SAC	*..sacrum*
PEL_ILCL	*..left iliac crista*
PEL_ILCR	*..right iliac crista*
PEL_TROCHL	*...left greater trochanter*
PEL_TROCHR	*...right greater trochanter*

Lower limb markers:

R/LKNE_L	*..right/ left lateral femoral epicondyle*
R/LKNE_M	*..right/ left medial femoral epicondyle*
R/LTH_1, 2, 3, 4	*..right/ left 4 thigh markers on a cluster*
R/LANK_L	*..right/ left lateral malleolus*
R/LANK_M	*..right/ left medial malleolus*
R/LSHA_1, 2, 3, 4	*..right/ left 4 shin markers on a cluster*
R/LFOOT_H	*..right/ left heal*
R/LFOOT_1H	*..right/ left head of the 1st metatarsal bone*
R/LFOOT_5H	*..right/ left head of the 5th metatarsal bone*
R/LFOOT_1B	*..right/ left base of the 1st metatarsal bone*
R/LFOOT_5B	*..right/ left base of the 5th metatarsal bone*
R/LFOOT_TOE	*..right/ left tip of 2nd toe*

Figure 1. The static calibration image as captured in Vicon with the list of all the markers used. Markers were placed on specific anatomical landmarks either directly on the skin or onto tight fitting elastic clothing. There were 49 individual markers placed, 16 markers attached on 4 marker clusters (with 4 markers per cluster) for lower extremities and 4 markers on an elastic head band. Palpation was used to identify bony landmarks and the same researcher placed the markers in order to minimize the inter-researcher variability. The 4 lower extremity clusters were fastened by an elasticated band/ wrap.

Protocol

A harness was fitted to participants, markers placed on the body as described above and a static subject calibration was recorded. Participants were then asked to ascend and descend the stairs in a step-over-step manner at a speed controlled by a metronome set at 90 beats per minute, since variables such as joint moments can be influenced by differences in walking speed[17]. This speed was selected as it has previously been shown to match closely the self-selected speed in elderly people during stair negotiation[18].

Participants were asked to start the ascent from the base of the staircase directly in front of the first step. At the top of the staircase they were asked to stop before turning around upon the researcher's instruction and position themselves at the edge of the top step in readiness for descending the stairs upon instruction. During the trial, participants were free to look anywhere they wanted, they were instructed not to hold onto the handrails unless needed in case of insecurity or instability and to try and match the given speed dictated by the metronome as closely as possible. Participants were asked to practise the tempo by marching on the spot a couple of times before the instruction to 'set off' was given by the researcher.

There were two conditions tested: 1. wearing the brace on the affected side (BR condition) and 2. the control condition with the brace off (CTR condition), the order of which was randomized prior to the start of the testing using a slip in a sealed envelope to limit any effect of possible fatigue or pain. The Bioskin® Q patellar tracking brace (Ossur UK Stockport, Manchester, UK). was used in the study (Figure 2). Three trials per condition were recorded, their mean was obtained and stance phases were analyzed for the affected leg in the conditions with and without the knee brace. No one used the handrails for the trials analysed.

BioSkin Q Brace Patella Tracking Brace

Figure 2. The Bioskin® Q patellar tracking brace (Ossur UK Stockport, Manchester, UK).

Data analysis

The kinematic and kinetic data recorded using the VICON system were labelled in VICON Nexus (Vicon Nexus 1.8.2). Post-processing calculation of the kinematic and kinetic data was conducted using Visual3D software (Visual3D Student Edition v4.96.9; C-Motion Inc., Rockville, MD, USA). All lower extremity segments were modelled as rigid bodies. Anatomical frames were defined by landmarks positioned at the medial and lateral borders of the joint, and for each segment a right handed co-ordinate systems was defined. Joint kinematics were calculated using an X–Y–Z Euler rotation sequence.

Joint kinetic data were calculated using three-dimensional inverse dynamics, and the exported internal joint moment data were normalised to body mass (Nm/kg). The internal joint moments correspond to external joint moments that have previously been reported in the literature - the internal knee extension moment corresponds to the external knee flexion moment and the internal knee abduction moment corresponds to the external knee adduction moment previously reported.

Statistical analysis was done using the Statistica software. Differences between the 2 conditions (BR and CTR) for all parameters were compared separately for both stair ascent and descent. The normality of distribution was tested using the Kolmogorov–Smirnov test and for the vast majority of variables the distribution was normal. Therefore a paired samples Student's *t*-test was used for the analysis. Although a correction for multiple comparisons was considered, there was a number of strong arguments against tests to correct for multiple comparisons[19,20]. Further, I did have an underlying rationale for specifically targeting the chosen dependent variables. For the reasons outlined above I have not adjusted the significance levels for multiple comparisons, however, I have presented results that are significant with specific alpha levels, thus enabling the reader to draw their own conclusions with regard to the statistical confidence and scientific relevance of the reported outcomes. Results in the text are presented as mean values with 95% confidence intervals (CI).

3.4 Results

Participant characteristics

There were 57.0 % female and 43.0 % male in the sample tested with the average age of 58.9 ± 7.7 years, height of 1.67 ± 0.10 m, body mass of 77 ± 15.9 kg and body mass index of 27.4 ± 3.8 kg/m^2. Participants had the following radiographic findings: The K-L grade 1/2/3 in the PF joint was 3.3/20.0/46.7 % of the sample, respectively. The K-L grade 1/2/3 in the TF joint was 6.7/20.0/43.3 % of the sample, respectively. The medial joint space narrowing (JSN) grade 1/2 was 30.0/40.0 % of the sample, respectively. The lateral JSN grade 0/1/2 was 13.3/50.0/6.7 % of the sample, respectively. The patellar JSN grade 1/2/3 was 43.3/20.0/6.7 % of the sample, respectively. Information about the remaining 17.0 % of the sample was documented by arthroscopy or MR imaging. All assessments were made by a consultant musculoskeletal radiologist.

Knee joint angles

During stair ascent, the brace significantly reduced the maximal knee flexion angle by 2.7 ^{0} [BR: 73.3^0, CI 71.8-74.7^0; CTR: 76.0^0, CI 74.0-78.0^0] and the total range of motion (ROM) at the knee by 2 ^{0} [BR: 61.1^0, CI 59.0-63.3^0; CTR: 63.1^0, CI 60.7-65.5^0] in the sagittal plane compared to the control condition. Additionally, the brace significantly reduced the maximal knee adduction (varus) angle by 2.1 ^{0} [BR: 4.9^0, CI 2.8-7.1^0; CTR: 7.0^0, CI 4.4-9.6^0] and the total ROM at the knee by 1.7 ^{0} [BR: 9.0^0, CI 7.0-10.9^0; CTR: 10.7^0, CI 8.8-12.6^0] in the frontal plane compared to the control condition. No significant differences were found in other measured knee joint angles (Figure 3, Table 1).

Table 1. Variables during the stance phase of stair ascent for brace and control conditions.

	Mean (95 % CI)	*p*

	Brace	Control	
	n = 30		
MIN Knee Flexion Angle (^{0})	12.1 (10.3-14.0)	12.9 (10.8-15.0)	*0.380*
MAX Knee Flexion Angle (^{0})	**73.3 (71.8-74.7)**	**76.0 (74.0-78.0)**	***0.002 †***
MAX Knee Abduction Angle (^{0})	4.0 (2.4-5.6)	3.7 (1.7-5.6)	*0.562*
MAX Knee Adduction Angle (^{0})	**4.9 (2.8-7.1)**	**7.0 (4.4-9.6)**	***0.044 †***
Total ROM in the Sagittal Plane (^{0})	**61.1 (59.0-63.3)**	**63.1 (60.7-65.5)**	***0.008 †***
Total ROM in the Frontal Plane (^{0})	**9.0 (7.0-10.9)**	**10.7 (8.8-12.6)**	***0.023 †***
Peak Knee Extension Moment (Nm/kg)	**1.00 (0.92-1.09)**	**1.05 (0.96-1.13)**	***0.043 †***
Peak Knee Abduction Moment (Nm/kg)	0.24 (0.18-0.30)	0.24 (0.18-0.30)	*0.947*
Gait speed (m/s)	0.50 (0.48-0.52)	0.50 (0.48-0.51)	*0.569*
Stride width (m)	0.11 (0.10-0.12)	0.11 (0.10-0.12)	*0.284*

† Significant difference between the two conditions ($p < 0.05$).

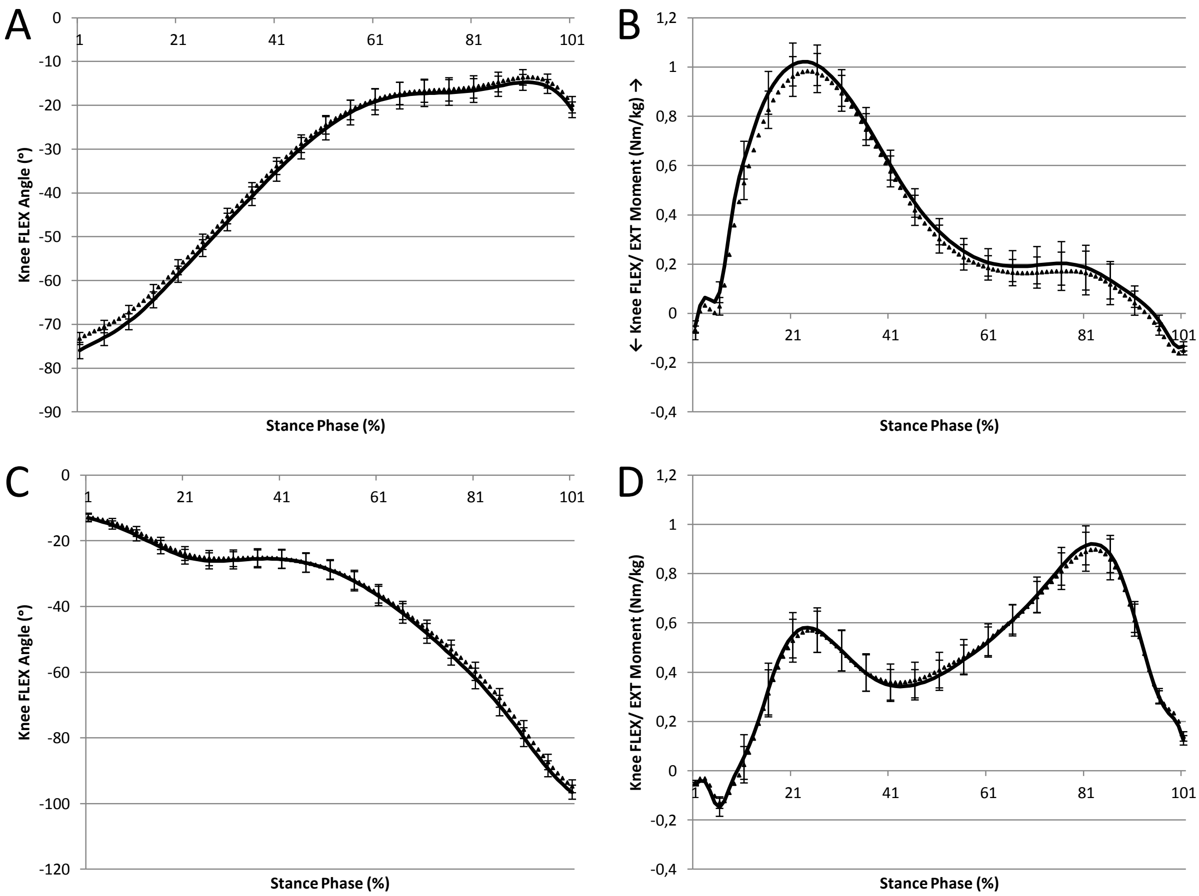


Figure 3. Mean knee joint angles and internal knee joint moments for the brace (∆) and control (—) conditions with 95% confidence intervals normalised to 100 % of stance phase at the standardised speed: sagittal angles during ascent (n = 30) (A), sagittal moments during ascent (n = 30) (B), sagittal angles during descent (n = 30) (C), sagittal moments during descent (n = 30) (D). The degrees of flexion on the y-axis (A, C) are reported as absolute values in the manuscript (tables).

During stair descent, the brace significantly reduced the maximal knee flexion angle by 1.8 ^{0} [BR: 94.7^0, CI 92.7-96.8^0; CTR: 96.5^0, CI 94.3-98.8^0] and the total ROM at the knee by 1.5 ^{0} [BR: 82.1^0, CI 80.1-84.0^0; CTR: 83.6^0, CI 81.5-85.7^0] in the sagittal plane compared to the control condition. No significant differences were found in other measured knee joint angles (Figure 3, Table 2).

Table 2. Variables during the stance phase of stair descent for brace and control conditions.

	Mean (95 % CI)		*p*
	Brace	Control	
	n = 30		
MIN Knee Flexion Angle (^{0})	12.7 (11.4-14.0)	13.0 (11.7-14.2)	*0.682*
MAX Knee Flexion Angle (^{0})	**94.7 (92.7-96.8)**	**96.5 (94.3-98.8)**	***0.039 †***
MAX Knee Abduction Angle (^{0})	3.4 (1.9-5.0)	3.0 (1.2-4.8)	*0.509*
MAX Knee Adduction Angle (^{0})	5.6 (3.8-7.5)	5.9 (3.7-8.0)	*0.812*
Total ROM in the Sagittal Plane (^{0})	**82.1 (80.1-84.0)**	**83.6 (81.5-85.7)**	***0.045 †***
Total ROM in the Frontal Plane (^{0})	9.1 (7.4-10.7)	8.9 (7.2-10.6)	*0.828*
Peak Knee Extension Moment (Nm/kg)	0.94 (0.86-1.02)	0.96 (0.88-1.05)	*0.209*
Peak Knee Abduction Moment (Nm/kg)	0.39 (0.30-0.47)	0.41 (0.32-0.49)	*0.145*
Gait Speed (m/s)	0.50 (0.48-0.52)	0.49 (0.47-0.51)	*0.313*
Stride Width (m)	0.15 (0.14-0.16)	0.15 (0.14-0.16)	*0.803*

† Significant difference between the two conditions ($p < 0.05$).

Knee joint moments

During stair ascent, the brace significantly reduced the peak internal knee extension moment by approximately 5 % [BR: 1.00 Nm/kg, CI 0.92-1.09 Nm/kg; CTR: 1.05 Nm/kg, CI 0.96-1.13 Nm/kg] compared to the control condition. No significant differences were found in the peak internal knee abduction moment (Figure 3, Table 1).

During stair descent, no significant differences were found between the two conditions when comparing the peak knee joint moments in the sagittal or frontal planes (Figure 3, Table 2).

Gait speed and stride width

No significant differences were found between the two conditions when comparing the gait speed or stride width during stair ascent and stair descent (Table 1 and 2).

3.5 Discussion

This is the first study demonstrating that a PF brace alters the biomechanics at the knee joint during stair negotiation in patients with predominantly PFOA (despite the fact that the results were small in magnitude) after it has been reported that PF braces can alleviate symptoms associated with PFOA[4,11,15]. During stair ascent, the brace reduced the knee flexion angle, the total knee ROM in the sagittal plane and the peak internal knee extension moment during the stance phase. In particular, the finding of a reduced peak internal knee extension moment with use of the brace would be expected to reduce PF loading. Interestingly, despite this brace not exerting any known mechanical effect in the frontal plane, as for example a valgus brace would do, the knee adduction (varus) angle and the total knee ROM in the frontal plane during the stance phase were reduced during stair ascent (but not during stair descent) while wearing the brace compared to the control condition. During stair descent the brace reduced the knee flexion angle and the total sagittal plane knee ROM during the stance phase, but it did not affect the internal knee extension moment, presumably not leading to any reduction in PF loading during stair descent unlike during stair ascent. However, when making inferences regarding internal joint loading (contact forces) based on internal joint moments, the limitations of inverse dynamics should be acknowledged in terms of the failure to account for muscle co-activation. However, this would only affect the present findings if there were marked differences in muscle co-activation between braced and control conditions. Nevertheless, it is important to note the magnitudes of the changes found in the present study with the brace on compared to no brace and critically evaluate what functional relevance these changes might bring to the OA patients. The 5 % reduction of the peak internal knee extension moment during stair ascent found in the present study might remain functionally irrelevant, and so the main benefit of the brace continues to be the pain-relieving effect found by our collaborators[11,15].

Effects of brace use

The maximal knee flexion angle was significantly reduced with the brace on at the beginning of the stance phase during stair ascent, when the participants were loading the stance leg while bringing the body up and onto the next step. This angle reduction in the sagittal plane is reflected by a reduced peak internal knee extension moment at the knee, possibly reducing the work required by the knee extensors and reducing PF loading. Although not reflected by any significant change in the peak moments in the frontal plane, the brace did reduce the maximal knee adduction angle during the beginning of the stance phase, which might have offered the participants a better mechanical alignment, stability and support compared to the control condition. During stair descent, the brace reduced the maximal knee flexion angle at the end of the stance phase, allowing the participants to bend their knees less while leaving the step compared to the control condition.

The results of the present study confirm the hypothesis that PF brace use can lead to alterations in knee kinematics and reductions in the joint moments about the knee predominantly in the sagittal plane, however, as I pointed out earlier these changes were minimal (a 5 % reduction in the peak internal knee extension moment during stair ascent with the brace on vs. no brace). It was speculated that the mechanism explaining these changes may be related to a greater perception of joint stability with use of the brace. This brace has previously been shown to cause reductions in PF pain when worn and assessed over a longer period[11,15], which may be at least partly attributed to the biomechanical changes shown here. In Study 1 (Chapter 2), I found that a knee OA group negotiated stairs with reduced sagittal knee joint moments compared to controls without knee OA[21]. The findings from the present study therefore suggest that this particular PF brace reduces the sagittal knee moment even further during stair ascent reducing the PF load. The functional relevance of these changes need to be investigated further to examine if these lead to actual benefits for the patients.

As the participants in the present study had also TFOA apart from PFOA, which was their predominant condition, I have investigated frontal plane kinematics and kinetics together with the sagittal plane biomechanics discussed above. In the present study, I found no change to the internal knee abduction moment with the brace during stair ascent or descent, which was to be expected since there is no mechanical function to this particular knee brace in contrast to a valgus brace for example, which exerts forces in the frontal plane. There was, however, a significant reduction in the knee adduction angle and the ROM in the frontal plane during stair ascent with brace use. As mentioned above, it was speculated that these kinematic changes could result from a greater perception of joint stability with use of the brace and be related to a generally “stiffer” gait observed in the OA participants. Although the brace did lead to small yet significant changes in sagittal (kinematic and kinetic) and frontal (kinematic) plane, the magnitude of these changes remains very small and so their clinical importance questionable.

Based on data from Study 1 (Chapter 2) as part of my PhD thesis, I can further provide reference values of age-, weight- and BMI-matched healthy controls without knee OA in addition to the present study findings in knee OA patients. During stair ascent the brace reduced the maximal flexion angle [brace 73.3 ± 3.9^0, no brace 76.0 ± 5.3^0, reference value healthy controls 78.6 ± 5.4^0], maximal knee adduction angle [brace 4.9 ± 5.8^0, no brace 7.0 ± 6.9^0, reference value healthy controls 4.0 ± 5.7^0], total ROM in the sagittal plane [brace 61.1 ± 5.7^0, no brace 63.1 ± 6.3^0, reference value healthy controls 63.1 ± 5.4^0], total ROM in the frontal plane [brace 9.0 ± 5.2^0, no brace 10.7 ± 5.1^0, reference value healthy controls 10.2 ± 4.4^0] and peak knee extension moment [brace 1.00 ± 0.23 Nm/kg, no brace 1.05 ± 0.23 Nm/kg, reference value healthy controls 1.35 ± 0.20 Nm/kg]. During stair descent the brace reduced the maximal flexion angle [brace 94.7 ± 5.6^0, no brace 96.5 ± 6.1^0, reference value healthy controls 98.1 ± 5.1^0] and total ROM in the sagittal plane [brace 82.1 ± 5.3^0, no brace 83.6 ± 5.6^0, reference value healthy controls 81.7 ± 5.1^0].

The standard speed was introduced because some of the key parameters I report can be altered by gait speed[17] and with the repeated measures design it was deemed important to ensure this potentially confounding variable was controlled. It was my intention to investigate whether the brace had any effect independent of any potential influence of gait speed. I was successful in matching gait speed between conditions allowing a valid comparison of the effect of the brace on the kinetic variables examined. Although I did not test the effect of the brace on walking speed in the present study, based on a study in patellofemoral pain patients during stair negotiation at self-selected speed, the brace does not change the walking speed or stance time[22].

Stride width was investigated to explain any possible differences between knee joint angles and moments (particularly in the frontal plane) as a result of wearing the brace.

However, no significant differences in stride width were found between conditions during stair ascent and descent.

The present study is novel with regard to knee OA, stair negotiation and PF bracing. Few studies have investigated the biomechanical effects of a brace as opposed to purely its effects on symptomatic pain relief. A recent study found that a PF Bio Skin Q brace, similar to the brace used in the present study, changed patellar kinematics in people with lateral PFOA during static postures in the unloaded and loaded knee, but these changes were not large enough to be clinically meaningful primarily because no reduction in pain was observed in their parent study[13]. In the present study participants chose themselves to use the realigning strap that is supplied with the brace based on their preferences. A randomized control trial has assessed the difference between using this brace (the Bio Skin Q Brace) with and without use of the strap in PFOA patients. They found no differences between the two conditions leading to the conclusion that the application of the strap did not make a difference in reducing the pain or PF joint stress[23]. They hypothesised that the brace alone without use of the strap increased the contact area and reduced the symptoms.

Limitations and recommendations

In the present study I did not examine or exclude participants with patella malalignment, which is associated with PFOA progression[24]. Similarly, I did not examine or exclude participants with varus or valgus alignment, and it is known that malalignment plays an important role when it comes to joint loading measures, such as the external knee adduction moment[25,26]. On the other hand, these were participants with predominantly PFOA and not exclusively medial compartment TFOA, which is associated with varus malalignment. This may slightly complicate the interpretation of my findings and possible recommendations for biomechanical interventions targeting specific knee compartments. However, such a group represents the knee OA population at large and brings predominant PFOA to closer attention of the scientific community. Lastly, these participants had experience of wearing this specific brace and therefore this was not an immediate assessment of the brace effect. This might have contributed to the relatively subtle differences found between the two conditions as the participants have adapted to wearing the brace over a period of time. Nevertheless, despite prior habituation to the brace I have observed biomechanical differences at the knee joint with its use.

As it is known from the literature and as mentioned before, the end-stage treatment for knee OA is total knee replacement[4]. Therefore efforts should be directed towards finding effective and cost-effective conservative treatment strategies such as knee bracing. However, given the large number of knee braces commercially available, it is important to evaluate them separately for their potential effects[23]. Additionally, other options remaining to be further investigated during ADL conditions in this population include taping, handrail or walking aid use, different walking/stair negotiation strategies and other orthotic devices. These various biomechanical interventions together with exercise, weight loss and education should remain at the centre of all therapeutic interventions in a knee OA population[27].

Conclusion

In summary, in a predominantly PFOA population it was found that a PF knee brace led to small yet significant changes in knee joint angles and moments, mainly in the sagittal plane, but also to a lesser extent in the frontal plane during stair ascent and descent.

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Chapter 4

Is stair handrail use an effective strategy to reduce knee joint loads in osteoarthritis patients?

Under review with Osteoarthritis and Cartilage

4.1 Abstract

Objective: The aim of this study was to investigate the effects of handrail use on knee biomechanics during stair negotiation in patients with knee osteoarthritis (OA).

Methods: Thirty participants with knee OA (40-70 years) negotiated a 7-step staircase under three conditions: 1. handrail on the affected side, 2. handrail on the contralateral side to the affected knee, and 3. control (CTR) condition without handrail. Kinematic data were obtained using motion analysis and ground reaction forces (GRF) were measured from force platforms embedded into the steps. Joint moments were calculated through inverse dynamics by combining kinematic and GRF data. Differences between conditions were compared separately for stair ascent and stair descent using a repeated measures analysis of variance.

Results: During stair ascent, the affected and contralateral handrail conditions reduced the peak knee extension moment of the affected leg by ~4 % compared to the CTR condition ($p < 0.001$). During stair descent, the contralateral handrail condition reduced the peak knee abduction moment of the affected leg by 15 % and 12 % compared to the CTR and affected conditions, respectively ($p < 0.001$). A narrower stride width was associated with handrail use during both stair ascent and descent compared to the CTR condition ($p < 0.005$).

Conclusion: The results show how handrail use alters knee joint loads in knee OA patients during stair negotiation, with implications for prevention of the mechanisms underpinning disease progression in patellofemoral and tibiofemoral knee OA.

4.2 Introduction

Osteoarthritis (OA) is the most common form of arthritis. The prevalence of the disease increases with age and as the global population is ageing, its management represents a growing socioeconomic burden[1]. Knee OA is one of the most prevalent forms of OA with around 12 % of Americans over 60 years and up to 10 % of British adults above 55 years suffering from symptomatic knee OA[2,3]. The healthcare costs are high due to the fact that there is no definitive cure for the disease and the treatment for end-stage disease is joint replacement[4]. Therefore, the need arises for simple and effective intervention strategies for improved management of the condition and/or for slowing disease progression.

In the recent past, the focus of biomechanics research has been on tibiofemoral (TF) OA with the majority of studies specifically investigating factors affecting the progression of medial TFOA. This reflects the fact that the medial compartment is more commonly affected than the lateral[2]. However, patellofemoral (PF) OA has now been recognised as an important sub-group that is more likely than TFOA to result in symptoms such as pain, stiffness and functional limitations impacting activities of daily living (ADL)[4-6];. Moreover, OA typically affects more than one compartment of the knee at the same time, resulting in a mixed disease of TFOA and PFOA, which represents a common clinical presentation. It has been shown that increased loading in the medial TF compartment characterised by a larger external knee adduction moment leads to TFOA progression[7,8]. In the case of PFOA, it is known that patella malalignment leads to PFOA progression[9] and it is expected that higher loading in the PF joint might also lead to pain and progression.

Biomechanical management strategies that are effective for TFOA are not necessarily effective for PFOA, since different mechanisms are responsible for their effects. It has been shown that altered PF loading leads to symptoms and structural progression. Therefore interventions to reduce the PF load seem appropriate to lessen symptoms and/or slow progression[10,11]. These interventions range from vastus medialis, glutei and trunk muscle retraining to orthotic devices such as foot orthoses and knee braces[11].

Stair negotiation in particular is known to be one of the most challenging ADL for people with knee OA. This is underpinned by findings of higher knee extension moments, higher medially directed ground reaction forces and higher external knee adduction moments during stair negotiation compared to level walking[12,13]. In people with PFOA, stair negotiation is considered the most common activity causing pain[14]. Additionally, there appears to be a natural strategy in patients with either isolated PFOA or a mixed PFOA and TFOA to reduce the internal knee extension moment and therefore reduce the PF loading compared to a control group during stair negotiation by flexing the knee less/ stiffening the leg[15,16] (Study 1/ Chapter 2). Therefore, other novel strategies to help reduce the joint moments and minimize joint loading even further are needed for effective management of the disease.

The use of walking sticks/canes during level walking has been tested as a potential strategy for unloading the painful joint. Research during level walking has shown benefits of contralateral walking stick use to the affected knee during level walking[17-19] by reducing medial knee loading, whereas the use of a walking stick on the same side as the affected knee actually increased medial knee loading[17]. It was hypothesised that using the handrails during ascending and descending stairs might have a similar impact on medial knee loading to that of using a walking stick on level ground. Additionally, it may also improve the ability for people with knee OA to negotiate stairs and enhance safety. For example, in older adults without knee OA, the use of handrails gave the participants additional perception of stability[20]. Light bilateral handrail use in healthy older adults has been shown to alter sagittal plane knee joint loading during stair ascent and descent compared to a condition without handrail use[21]. Currently, the effects of handrail use on knee loading in OA remains unknown, but is a very pertinent issue to investigate due to the markedly higher loads present during stair negotiation compared to level walking as mentioned above. If the previously reported findings of reduced sagittal plane moments in healthy older participants with handrail use[21] also hold true for a population with knee OA, this could be a beneficial strategy particularly for PFOA.

The aim of this study was, therefore, to investigate the effects of handrail use on knee joint biomechanics during stair negotiation in individuals with mixed knee OA but predominant PF symptoms. It was hypothesised that the handrail use would reduce frontal plane knee moments with implications for TFOA and also reduce sagittal plane knee moments with implications for PFOA.

4.3 Methods

Participants

Thirty participants were recruited from local primary care centres, hospital-based orthopaedic, rheumatology and physiotherapy clinics. Participants had mixed knee OA with predominantly PF symptoms but also medial TF involvement. Ethical approval was obtained from the relevant bodies and written informed consent was obtained from all participants.

Inclusion criteria

Participants were included if they were aged between 40-70 years, had a Kellgren-Lawrence (K-L) score of grade 2 or 3 in the PF joint, greater than the K-L score for the TF joint of the same knee. Participants needed to have PF joint symptoms such as pain reproduced with stair climbing, kneeling, prolonged sitting or squatting and lateral or medial patellar facet tenderness on palpation or a positive patellar compression test (patient contracts their quadriceps fully whilst their patella is compressed against the femur by the assessor). Pain must have been present daily for the previous 3 months and rated equal to or above a score of 40 on a 0-100-mm Visual Analogue Scale (VAS) on the day of the assessment for a nominated aggravating activity, which usually involved stairs or prolonged sitting. Some patients (n = 10) also had symptomatic knee OA in the contralateral leg but this was confirmed to be of a much lesser severity compared to the affected side.

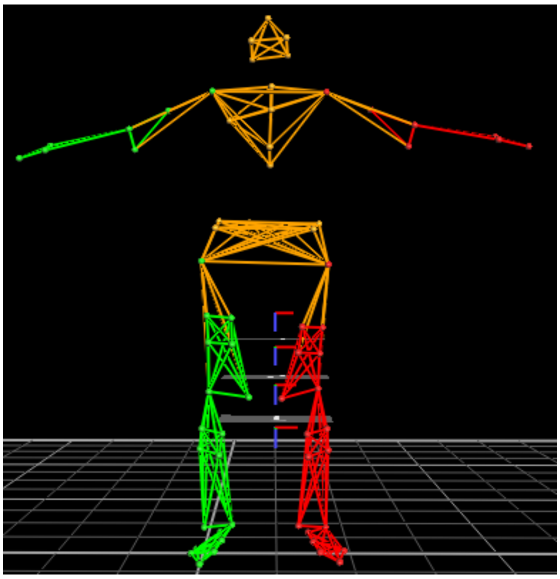
Exclusion criteria

Participants were excluded if the predominant symptoms emanated clinically from the TF joint from meniscal or ligament injury, if they had rheumatoid arthritis or other forms of inflammatory arthritis or if they had received an intra-articular steroid injection into the painful knee in the previous month. All radiographs were examined by a consultant musculoskeletal radiologist. Clinical assessments were performed by an experienced musculoskeletal physiotherapist.

Experimental set up

A seven-step staircase instrumented with 4 individual force plates (Kistler, Winterthur, Switzerland; recording at 1000 Hz, 300 x 500 mm), embedded into the second, third, fourth and fifth steps, respectively, was used in the study. The step dimensions represented standard stair dimensions with a going of 275 mm, a riser height of 175 mm and a width of 1050 mm. Handrails were present on both sides at a height of 900 mm above the steps and at an inclination angle of 31°. All participants wore a full body safety harness attached via a rope to an overhead safety rail and a belaying assistant on the side of the stairs. This procedure during the stair testing was a safety precaution to support the participant in the case of a fall during the tests. The belaying was conducted carefully so as not to interfere with the natural stair gait.

Kinetic and kinematic data were collected using the force plates and a 10-camera VICON T20 optoelectronic motion analysis system (VICON motion systems Ltd., Oxford, UK; recording at 100 Hz), tracking a set of 69 passive retro-reflective markers using a modified 6 Degrees of freedom (6 DoF) whole-body model developed by Capozzo and colleagues (Figure 1)[22].



Head markers:

- elastic head band with 4 markers

HEAD_BR	*..back right*
HEAD_FR	*..front right*
HEAD_BL	*..back left*
HEAD_FL	*..front left*

- 1 marker for static calibration only

HEAD_TOP	*..head top*

Trunk markers:

TOR_C7	*..7th cervical vertebra*
TOR_T10	*..10th thoracic vertebra*
TOR_CLAV	*..jugular notch*
TOR_STER	*..xiphoid process*
SCAPI	*..left scapula*
TOR_SHR/L	*..right/ left acromion*

Upper limb markers:

R/LARM_ELL	*..right/ left lateral elbow*
R/LARM_UP	*..right/ left middle upper arm*
R/LARM_ELM	*..right/ left medial elbow*
R/LARM_WRU	*..right/ left ulnar wrist*
R/LARM_WRR	*..right/ left radial wrist*
R/LARM_FIN	*..right/ left head of 2^{nd} metacarpal bone*

Pelvic markers:

PEL_PSISL	*..left posterior superior iliac spine*
PEL_ASISL	*..left anterior superior iliac spine*
PEL_PSISR	*..right posterior superior iliac spine*
PEL_ASISR	*..right anterior superior iliac spine*
PEL_SAC	*..sacrum*
PEL_ILCL	*..left iliac crista*
PEL_ILCR	*..right iliac crista*
PEL_TROCHL	*...left greater trochanter*
PEL_TROCHR	*...right greater trochanter*

Lower limb markers:

R/LKNE_L	*..right/ left lateral femoral epicondyle*
R/LKNE_M	*..right/ left medial femoral epicondyle*
R/LTH_1, 2, 3, 4	*..right/ left 4 thigh markers on a cluster*
R/LANK_L	*..right/ left lateral malleolus*
R/LANK_M	*..right/ left medial malleolus*
R/LSHA_1, 2, 3, 4	*..right/ left 4 shin markers on a cluster*
R/LFOOT_H	*..right/ left heal*
R/LFOOT_1H	*..right/ left head of the 1^{st} metatarsal bone*
R/LFOOT_5H	*..right/ left head of the 5^{th} metatarsal bone*
R/LFOOT_1B	*..right/ left base of the 1^{st} metatarsal bone*
R/LFOOT_5B	*..right/ left base of the $5t^{h}$ metatarsal bone*
R/LFOOT_TOE	*..right/ left tip of 2^{nd} toe*

Figure 1. The static calibration image as captured in Vicon with the list of markers used.

Markers were placed on specific anatomical landmarks either directly on the skin or onto tight fitting elastic clothing. There were 49 individual markers placed, 16 markers attached on 4 marker clusters (with 4 markers per cluster) for lower extremities and 4 markers on an elastic head band. Palpation was used to identify bony landmarks and the same researcher placed the markers in order to minimize the inter-researcher variability. The 4 lower extremity clusters were fastened by an elasticated band/ wrap.

Protocol

A harness was fitted to participants, markers placed on the body as described above and a static participant calibration was recorded by the motion capture system. Participants were then asked to ascend and descend the stairs in a step-over-step manner (one foot striking each step) at a speed controlled by a metronome set at 90 beats per minute. Controlled speed was used to examine the effects of handral use independent of the speed and because joint moments can be influenced by differences in walking speed[23]. This speed was selected as it has previously been shown to match closely the self-selected speed in elderly people during stair negotiation[24].

Participants were asked to start the ascent from the base of the staircase directly in front of the first step and from the edge of the top step for the descent. During the trial, participants were instructed to try and match the given speed dictated by the metronome as closely as possible. Participants were asked to practise the tempo by marching on the spot several times before the instruction to 'set off' was given.

There were three conditions tested: 1. handrail use on the affected side (affected condition), 2. handrail use on the contralateral side to the affected knee (contralateral condition) and 3. the control (CTR) condition with no handrail use. The order of conditions was randomized using sealed envelopes to limit any possible order effects. Three trials per condition were recorded and stance phases were analyzed for the affected leg (or the more severely affected leg in case of bilateral disease) for all three conditions. With each foot striking the four force plates twice during stair ascent and descent, the mean of six stance phases (two stance phases per trial; three trials) was used for the analysis. During the CTR condition the participants were instructed not to hold onto the handrails unless absolutely needed in case of insecurity or instability; none of the participants used the handrails during this condition.

Data analysis

The kinematic and kinetic data recorded using the VICON system were labelled in VICON Nexus (Vicon Nexus 1.8.2). Post-processing of the kinematic and kinetic data was conducted using Visual3D software (Visual3D Student Edition v4.96.9; C-Motion Inc., Rockville, MD, USA). All lower extremity segments were modelled as rigid bodies. Anatomical frames were defined by landmarks positioned at the medial and lateral borders of the joint, and for each segment a right handed co-ordinate systems was defined. Joint kinematics were calculated using an X–Y–Z Euler rotation sequence. Joint kinetic data were calculated using three-dimensional inverse dynamics, and the exported internal joint moment data were normalised to body mass (Nm/kg). Pain during the task was assessed using a 0-100-mm VAS, on which the participants were asked to mark their level of knee pain during each condition for ascent and descent separately.

Statistical analysis

Statistical analysis was done using Statistica software. Differences between the three conditions for all parameters were compared separately for stair ascent and stair descent using a repeated measures analysis of variance (ANOVA) and Fisher´s Least Significant Difference test for post-hoc testing. The pain scores were tested for differences between the three conditions using Friedman ANOVA.

4.4 Results

Participant characteristics

There were 17 females (57 %) out of the 30 participants in the sample tested, with an average age for the whole sample of 58.9 ± 7.7 years, height of 1.67 ± 0.10 m, body mass of 77 ± 15.9 kg and body mass index of 27.4 ± 3.8 kg/m^2. Radiography was available for 21 participants and they had the following radiographic findings: The K-L grade 2/3 in the PF joint was 20/46.7 % of the sample, respectively. The K-L grade 2/3 in the TF joint was 20/43.3 % of the sample, respectively. Information on disease status in the remaining 9 participants (30 % of the sample) was documented from arthroscopy or MR imaging.

Effects of handrail use on knee joint moments

During stair ascent, handrail use reduced the peak sagittal plane knee joint moment, more specifically both the affected and contralateral condition significantly reduced the peak knee extension moment of the affected leg by ~4 % compared to the CTR condition. No differences between the three conditions were found in the peak frontal plane knee joint moment during stair ascent (Figure 2A and 2B, Table 1).

Table 1. Variables during the stance phase of **stair ascent** for control (CTR), contralateral (CL) and affected (AF) conditions.

	Mean (SD)	*p*

	CTR	CL	AF	*overall*
		n = 30		
Knee Moments (Nm/kg):				
Peak Extension	**1.05 (0.23)** †,*	**1.01 (0.23)** †	**1.01 (0.23) ***	***0.001***
Peak Abduction	0.24 (0.17)	0.25 (0.16)	0.24 (0.17)	*0.490*
Knee Angles (^{0}):				
MIN Flexion	**12.9 (5.6) ***	**13.2 (5.6)** ††	**11.9 (5.7)** *,††	***0.014***
MAX Flexion	**76.0 (5.3)** †,*	**74.9 (5.8)** †	**74.8 (5.5) ***	***0.010***
MAX Abduction	3.7 (5.2)	3.9 (5.4)	3.8 (5.3)	*0.652*
MAX Adduction	7.0 (6.9)	7.2 (7.3)	7.4 (6.9)	*0.392*
Total Sagittal ROM	**63.1 (6.3)** †	**61.7 (6.2)** †,††	**62.9 (6.3)** ††	***0.029***
Total Frontal ROM	10.7 (5.1)	11.2 (5.5)	11.2 (5.3)	*0.184*
Other parameters:				
Gait speed (m/s)	0.50 (0.05)	0.50 (0.05)	0.49 (0.05)	*0.127*
Stride width (m)	**0.108 (0.024)** †,*	**0.102 (0.023)** †	**0.099 (0.019) ***	***0.003***
Pain (mm)	24 (24)	23 (23)	25 (24)	*0.475*

Data are presented as mean (SD).

Significant difference between groups ($p < 0.05$):

†: CTR vs. CL

*: CTR vs. AF

††: CL vs. AF

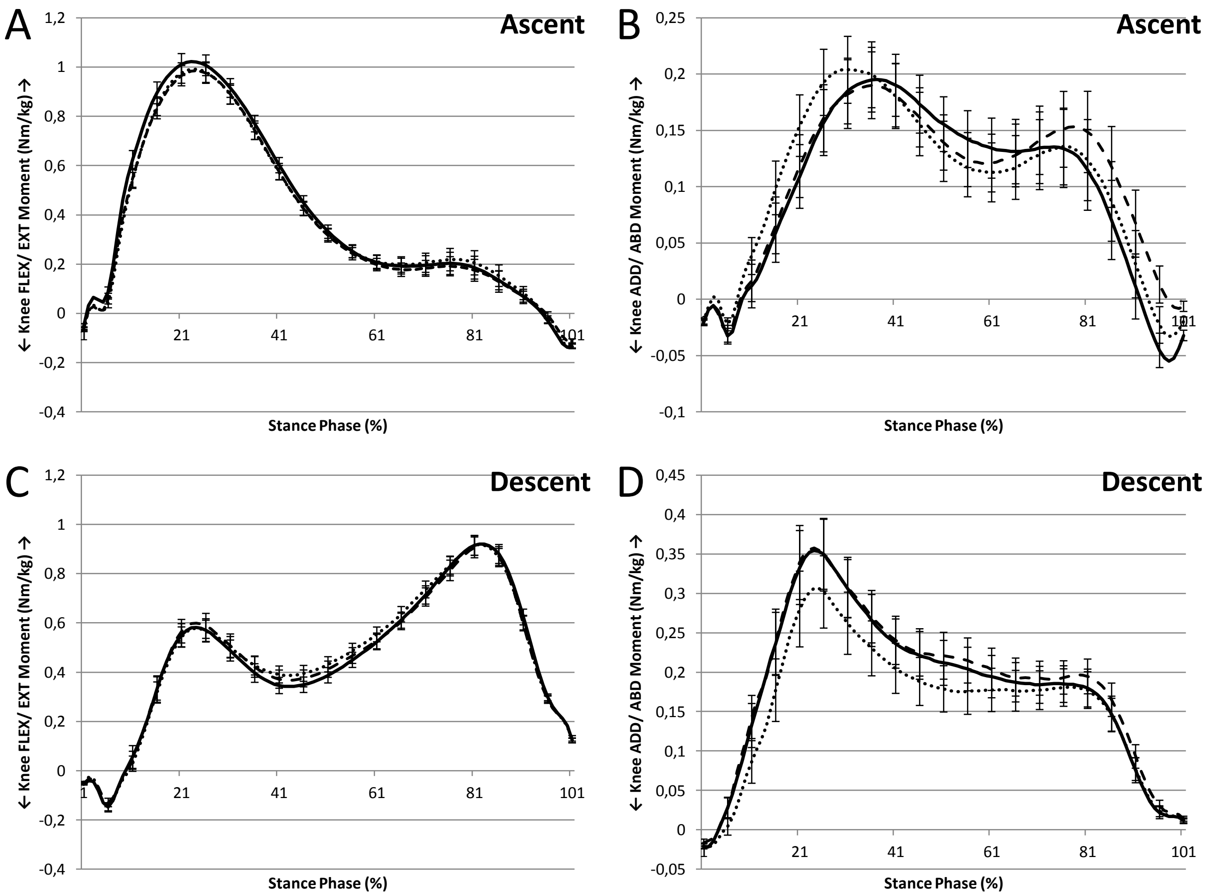


Figure 2. Mean internal knee joint moment for the control condition of no handrails (black line), contralateral condition (dotted) and handrail on the affected side condition (dashed) with standard error of the mean normalised to 100 % of stance phase at the standardised speed: sagittal knee joint moment during ascent (A), frontal knee joint moment during ascent (B), sagittal knee joint moment during descent (C), frontal knee joint moment during descent (D).

During stair descent, no differences between the three conditions were found in the peak sagittal plane knee joint moment. In the frontal plane, the contralateral condition significantly reduced the peak knee abduction moment of the affected leg by 15 % and 12 % compared to both the CTR and affected conditions, respectively. (Figure 2C and 2D, Table 2).

Table 2. Variables during the stance phase of **stair descent** for control (CTR), contralateral (CL) and affected (AF) conditions.

	Mean (SD)			*p*
	CTR	CL	AF	*overall*
		n = 30		
Knee Moments (Nm/kg):				
Peak Extension	0.96 (0.23)	0.95 (0.22)	0.95 (0.23)	*0.538*
Peak Abduction	**0.41 (0.23) †**	**0.35 (0.24) †, ††**	**0.40 (0.24) ††**	***0.000***
Knee Angles (^{0}):				
MIN Flexion	13.0 (3.3)	13.1 (3.6)	12.6 (3.8)	*0.127*
MAX Flexion	96.5 (6.1)	96.2 (6.1)	96.3 (5.9)	*0.534*
MAX Abduction	3.0 (4.8)	2.9 (5.1)	3.0 (4.7)	*0.855*
MAX Adduction	5.9 (5.8)	5.8 (6.0)	6.1 (5.6)	*0.384*
Total Sagittal ROM	83.6 (5.6)	83.0 (5.9)	83.7 (6.0)	*0.156*
Total Frontal ROM	8.9 (4.6)	8.7 (4.6)	9.1 (4.3)	*0.408*
Other parameters:				
Gait speed (m/s)	0.49 (0.06)	0.49 (0.05)	0.49 (0.06)	*0.516*
Stride width (m)	**0.149 (0.024) ***	0.144 (0.033)	**0.138 (0.026) ***	***0.005***
Pain (mm)	31 (29)	29 (27)	32 (30)	*0.138*

Data are presented as mean (SD).

Significant difference between groups ($p < 0.05$):

†: CTR vs. CL

*: CTR vs. AF

††: CL vs. AF

Effects of handrail use on knee joint angles

During stair ascent, handrail use altered sagittal plane kinematics, albeit with a relatively small effect. In particular, the affected condition significantly reduced the minimal knee flexion angle of the affected leg by 1.0 ^{0} and 1.3 ^{0} compared to the CTR and contralateral conditions, respectively. The maximal knee flexion angle of the afftected leg was significantly reduced by 1.1 ^{0} and 1.2 ^{0} in the contralateral and affected conditions, repectively, compared to the CTR condition. The total range of motion (ROM) of the affected leg in the sagittal plane was significantly reduced in the contralateral condition by 1.4 ^{0} and 1.2 ^{0} compared to the CTR and affected conditions, respectively. No differences were found in the frontal knee joint angles (Table 1). During stair descent, no differences were found between the three conditions in sagittal or frontal knee joint angles (Table 2).

Effects of handrail use on gait speed, stride width and pain

No differences were found between the three conditions when comparing the gait speed or pain during either stair ascent or stair descent. However, when comparing the stride width, a significantly narrower stride width was used during both handrail conditions compared to the CTR condition during stair ascent and the affected condition compared to the CTR condition during stair descent (Tables 1 and 2).

4.5 Discussion

This study demonstrates how handrail use on the affected side and contralateral side alters the biomechanics at the knee joint during stair negotiation in a mixed knee OA population with predominant PFOA symptoms. The results of the present study confirm the hypothesis that handrail use can indeed lead to alterations in knee joint loading by reducing the joint moments about the knee during stair negotiation in both sagittal and frontal planes, impacting the mechanisms responsible for disease progression in both PFOA and TFOA. The nature of these changes was different for stair ascent and descent and will be discussed further in the following sections.

Implications of handrail use for PFOA

During stair ascent, handrail use on either the affected or contralateral side reduced the sagittal plane knee joint moment compared to the unaided CTR condition. This finding of a reduced sagittal plane moment with the use of handrails on either side is important as it is likely to reduce loading in the painful PF compartment and could therefore be a simple yet effective strategy to mitigate disease progression. Knee kinematics during stair ascent were affected by handrail use, but to a lesser extent than the joint moments. Therefore using the handrails on either side could be beneficial in PFOA and for mixed knee OA with predominant PFOA (such as the present sample) to reduce the PF loading via reductions of the knee extension moment. In my first study (Chapter 2), I found that the OA group negotiated stairs with a reduced sagittal knee joint moment compared to age-matched controls without knee OA[16]. Findings from the present study suggest that using the handrail on either side reduces the sagittal knee moment even further during stair ascent in this cohort, reducing the PF load even further. Interestingly, it was found that whilst bilateral handrail use reduced the ankle joint moment, it increased the peak knee extension moment during stair ascent compared to the unaided condition in thirteen healthy elderly (in contrast to the sample in the present study)[21]. Additionally, in the study with healthy elderly the participants were specifically asked to "lightly touch" rather than hold onto the handrails[21], whereas in the present study the participants were free to use them more heavily if they wished. Although the forces applied on the handrails were not quantified in either study, they were most likely greater in the present study compared to the previous, which may at least in part explain the differences. The idea to use a handrail during stair negotiation as a means of support, stems from research done on healthy individuals and in a knee OA population during level walking with the use of a walking stick. For example, my findings during stair negotiation are similar to results of another study that reported that walking stick use on either ipsilateral (on the afftected side) or contralateral side to the study leg unloaded the limb in healthy young adults during level walking[25].

In contrast to stair ascent, during stair descent I found no differences between the three conditions in the sagittal plane knee joint kinetics or kinematics. Several potential mechanisms may explain this finding. Firstly, the sagittal knee joint moment during stair descent was lower (regardless of condition) compared to ascent and therefore participants may not have needed to reduce the loading as much as during stair ascent. Secondly, participants may have found it less practical to use the handrails to unload the lower limbs during descent compared to ascent. However, this is a speculation as handrails in the present study were not instrumented. Thirdly, Figure 2 indicates that the peak sagittal moment during descent occured when the leg was supporting on the step above and the leading leg has already contacted the step below, hence, much support from the handrails may not have been required at this stage of the stance phase.

Implications of handrail use for TFOA

During stair descent, handrail use on the contralateral side reduced the peak knee abduction moment compared to the CTR and affected conditions. Therefore, this strategy led to a reduction of the medial knee loading, presumably by aligning the body centre of mass more directly over the stance leg. This adaptation must have reduced the lever arm of the ground reaction force in the frontal plane around the knee joint, to reduce the moment. The present study findings are therefore in line with the reduced frontal plane knee joint moment reported with contralateral walking stick use during level walking[18,19]. No change in the frontal plane knee joint moment and hence no alteration to the loading on the medial knee compartment was observed during the affected condition compared to CTR. Therefore, contralateral handrail use on stairs similarly to contralateral walking stick use on level ground leads to unloading of the medial knee compartment. Contralateral handrail use on stairs could therefore be recommended as a strategy for patients with either medial TFOA, or mixed disease with medial TF involvement such as my sample to mitigate disease progression. There were no differences between the three conditions in the frontal plane knee joint kinematics during stair descent. In terms of the research on walking stick use, it was found that contralateral walking stick use led to a 10 % reduction of the external knee adduction moment in people with TFOA during level walking compared to no walking stick[18]. In comparison, on stairs I found a 15 % and 12 % reduction of the knee abduction moment by contralateral handrail use compared to the CTR and affected condition, respectively. One study reported that the more body weight goes through the walking stick, the greater the external knee adduction moment reduction in the contralateral knee during level walking[19]. Another study looked at both ipsilateral (on the affected side) and contralateral walking stick use during level walking in females with TFOA and found that contralateral cane use was recommended as it produced the lowest peak knee abduction moment. No walking stick was found to be preferable over ipsilateral walking stick use as the ipsilateral use of the walking stick led to highest knee moment produced out of all conditions[17]. On stairs, another possibly effective strategy could be bilateral handrail use imitating bilateral walking aid use. Research on bilateral walking pole use led to a 27 % reduction in medial contact force using in vivo contact force data in a single subject after knee replacement. Furthermore, it caused 11 % and 21 % reductions in the lateral and total contact force, respectively[26]. In contrast, another study found that bilateral walking pole use increased the knee adduction moment and therefore did not reduce the medial load at the knee in people with medial TFOA during level walking[27]. However, replicating bilateral walking pole use with handrails is perhaps not very meaningful, as patients might not have such a setting available during ADL, i.e., a handrail typically is present on one side only. Whether these findings of bilateral walking stick aided level ground walking use translate to stair negotiation and bilateral handrail use remain to be confirmed by future research.

During stair ascent, there were no differences between the three conditions in frontal plane knee joint kinetics or kinematics. One of the possible explanations for why I did not observe a reduction in the frontal plane knee joint moment with contralateral handrail use during stair ascent as I did during stair descent is that the magnitude of the frontal plane knee joint moment was higher during stair descent compared to that during ascent. Therefore, participants may have used this strategy of contralateral handrail use to reduce the loading during the condition where it was needed the most, i.e., during stair descent where the frontal plane knee joint moments were the highest.

The controlled speed was used as some of the key parameters I report can be altered by gait speed[23]. I was successful in matching gait speed between conditions allowing a valid comparison of the effect of the handrail on the variables examined. There was no difference in pain levels across the three conditions during either stair ascent or descent. I hypothesised that it might have been due to the fact that the participants were in some level of pain upon arrival and throughout the testing. Stair negotiation is a challenging and painful activity for these patients[12,14] and so it is possible that they simply were in pain and could not really differentiate whether one condition was slightly better in terms of pain than the other.

Limitations and recommendations

In the present study, I neither examined nor excluded participants with varus or valgus malalignment, which plays an important role in measuring joint loading[28,29]. However, the participants had predominantly PFOA and not exlusively medial TFOA, which is associated with varus malalignment. Secondly, the handrails were not instrumented and so I was unaware how much force was being transmitted and how much the participants used the rails to help themselves during stair ascent and descent. Nevertheless, I am confident that what I measured from the ground reaction forces (and therefore calculated for the joint moments) resulted from my experimental manipulation of handrail use. The end-stage treatment for knee OA is total knee replacement and therefore, efforts should be directed towards finding simple, effective and accessible conservative treatment strategies such as the use of walking sticks and stair handrails.

Conclusion

In summary, I present novel data to show that in a mixed knee OA population handrail use is an effective strategy for favourably modifying knee joint loads during stair negotiation for both medial TFOA and PFOA. Handrail use on either the affected or contralateral sides is recommended for PFOA, whilst contralateral handrail use is recommended for medial TFOA.

4.6 References

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Chapter 5

Does knee osteoarthritis impair balance during stair negotiation?

Under review with Journal of Biomechanics

5.1 Abstract

Objective: The aim of the study was to investigate if knee OA patients display balance impairments during stair negotiation and whether contralateral handrail use improves balance control.

Methods: Thirty participants (58.9 ± 7.7 years) with mixed knee OA and thirty age- and BMI-matched controls negotiated a 7-step staircase with ground reaction forces measured from force platforms embedded into steps. Kinematic data were obtained using a 10-camera motion analysis system tracking a full body marker set. Balance was quantified by measuring the separation between the centre of mass (CoM) and centre of pressure (CoP) in mediolateral (M-L) and anteroposterior (A-P) planes. The CoM velocity was measured in the same planes. Pain was assessed using a visual analogue scale.

Results: The OA group experienced more pain compared to controls and there were no differences between groups in CoM-CoP separation during stair negotiation. The OA group descended stairs with a greater peak vertical downward CoM velocity *($p=0.028$)*, stride width *($p=0.043$)* and slower gait speed *($p=0.009$)* compared to controls. During stair ascent handrail use in the OA group reduced the mean M-L CoM velocity *($p<0.001$)*, stride width *($p=0.023$)* and increased the maximum posterior CoM-CoP separation *($p=0.008$)*. During stair descent, handrail use in the OA group reduced the mean M-L CoM velocity *($p<0.001$)* and range of the M-L CoM-CoP separation *($p=0.032$)*.

Conclusion: No major balance impairments were identified in knee OA patients during stair negotiation, although they had greater difficulty during stair descent. The use of contralateral handrail had a minimal effect on balance control.

5.2 Introduction

Osteoarthritis (OA) is a chronic degenerative disease affecting joints. The prevalence of OA increases with age and presents a growing socioeconomic burden as the global population is ageing. As a load-bearing joint, the knee is one of the most commonly affected[1-3]. It has been estimated that around 12 % of the US population over 60 years and 10 % of the British population above 55 years suffer from symptomatic knee OA[4,5]. Previous biomechanics research in knee OA has predominantly focused on tibiofemoral (TF) OA, but recently patellofemoral (PF) OA has received more attention from biomechanical investigations and interventions. PFOA is more likely to result in symptoms such as pain, stiffness and functional limitations impacting activities of daily life (ADL) compared to the TFOA[6-8]. From a clinical perspective, knee OA typically affects multiple compartments at the same time resulting in a mixed disease of TF and PFOA.

Stair negotiation presents a particular problem for knee OA patients[9], causing pain[10] and likely posing higher biomechanical demands on the joint during this task, with greater frontal plane knee joint moments and internal knee extension moments compared to level walking[9,11]. Interestingly, people with PFOA or mixed knee OA seem to adopt a strategy that leads to a reduction of the internal knee extension moment compared to a control group both during stair ascent and descent[12,13]. Furthermore, mixed knee OA patients ascend stairs with an increased peak internal knee abduction moment compared to controls[12] (Study 1/ Chapter 2).

The combination of pain and altered knee biomechanics in knee OA together with the high biomechanical demands during stair negotiation may lead to balance impairments. It is well established that measures of standing balance, for example by using a swaymeter, are poorer in knee OA patients compared to healthy controls[14]. Patients with moderate to severe OA seem to have greater deficits in static and dynamic balance control than those with mild disease using posturography and clinical tests such as single leg standing or functional tests such as the “Timed up and go“. These findings have been interpreted as translating to poorer balance performance in the knee OA population[15,16]. Duffell et al. (17) found that people with early TFOA had reduced postural balance on both their affected and unaffected limbs during single leg standing. During stair negotiation, patients with knee OA demonstrated less time in single support and decreased total gait velocity compared with controls[18]. However, an understanding of the effects of knee OA on balance during gait activities and specifically during stair negotiation is missing.

A robust ways to assess balance during gait is by investigating the separation between the centre of mass (CoM) and the centre of pressure (CoP). The CoM-CoP separation indicates dynamic balance during gait; the greater the separation, the greater the challenge to maintain balance[19]; a more lateral (or anterior) position of the CoM in relation to the CoP (i.e. greater CoM-CoP separation) leads to a more unstable body position in lateral (or anterior) direction. The largest CoM-CoP separation representing positional instability occurs towards the end of the single leg stance phase[20]. Additionally, the velocity of the CoM can provide unique insights to the level of “control” of dynamic balance during stair negotiation[21]. A lower CoM velocity might indicate more control over the CoM, whereas a higher CoM velocity in certain phases of the gait cycle may indicate a reduced control over the CoM (and therefore impaired balance control). For example, Buckley et al. (21) found a reduced vertical downwards CoM velocity in healhty elderly compared to young possibly as a more cautious strategy to descend stairs in light of their reduced eccentric strength to absorb the impact associated with landing.

Handrail use has been shown to modify knee loading and the CoM-CoP separation during stair negotiation in older adults without knee OA[22]. During stair descent in particular, the use of handrails caused a redistribution of joint moments between the knee and ankle, while the peak ankle joint moment increased, the peak knee joint moment decreased. Also, the use of handrails compared to unaided stair descent led to a greater peak sagittal CoM-CoP separation with the CoM behind the CoP and smaller peak sagittal CoM-CoP separation with the CoM in front of the CoP[22]. I have recently shown that contralateral handrail use reduced the peak internal knee extension moment during stair ascent and the peak internal knee abduction moment during stair descent compared to not using the handrail in knee OA patients[23] (Study 3/ Chapter 4). Since handrail use has been shown to modify knee joint loading, it may be hypothesised to improve balance control in patients with knee OA. However, this currently remains unknown, and similarly a biomechanical analysis of the CoM velocity or CoM-CoP separation in knee OA population during stair negotiation is missing.

The aim of the study was to investigate whether the presence of knee OA compromised balance during stair negotiation and if so, whether handrail use could improve balance. I hypothesised that i) patients with knee OA would negotiate stairs with a higher CoM velocity and a larger CoM-CoP separation compared to controls and ii) that handrail use would lead to a reduced CoM velocity and a smaller CoM-CoP separation in people with knee OA, making stair negotiation more stable.

5.3 Methods

Participants

Thirty knee OA participants were recruited from local primary care centres. Thirty control (CTR) participants without knee OA matched for age and body mass index were recruited from the local region via retirement groups and university staff. Ethical approval was obtained from the relevant bodies and written informed consent was obtained from all participants.

Inclusion criteria

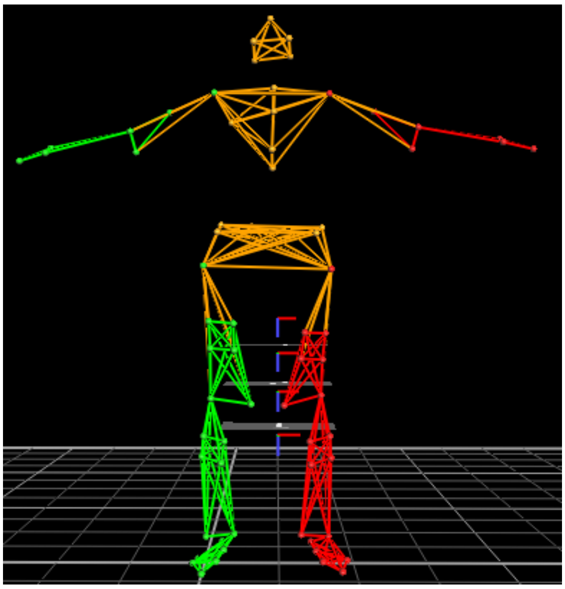
Participants were included in the OA group if they were aged between 40 -70 years. All participants had mixed compartment OA with a Kellgren-Lawrence (K-L) score grade 2 or 3 in the PF joint which was equal to or greater than the K-L score for the TF joint of the same knee. The diagnosis was made in 21 participants by plain radiography. Nine participants had arthroscopic or MR imaging documented evidence of the mixed disease severity and distribution. Pain must have been present daily for previous 3 months and rated a score equal to or above 40 on a 0-100-mm Visual Analogue Scale (VAS) on the day of the assessment for a nominated aggravating activity. In 10 out of the 30 knee OA participants there was radiographic and symptomatic evidence of bilateral disease. In these cases, the most symptomatic knee was chosen as the affected knee.

Exclusion criteria

Participants were excluded from the OA group if symptoms were traumatic in origin, if they had rheumatoid arthritis or other forms of inflammatory arthritis or if they had an intra-articular steroid injection into the painful knee in the previous month. Radiographs were read by a consultant musculoskeletal radiologist and clinical assessments were made by an experienced physiotherapist. Participants were excluded from the CTR group if they had knee pain assessed by the Knee Osteoarthritis Outcome Score questionnaire, lower extremity problems or any other issues affecting their gait, such as recent injuries, neurological conditions, etc.

Experimental set up

A seven-step staircase instrumented with 4 individual force plates (Kistler, Winterthur, Switzerland; recording at 1000 Hz, 300 x 500 mm), embedded into second, third, fourth and fifth steps, was used. Step dimensions represented standard dimensions with a going of 275 mm, riser height of 175 mm and width of 1050 mm. The handrails were at a height of 900 mm and an inclination of 31º. Participants wore a full body safety harness during the testing. Kinetic data were collected using force plates and kinematic data using a 10-camera VICON T20 optoelectronic motion analysis system (VICON motion systems Ltd., Oxford, UK; recording at 100 Hz). Both kinetic and kinematic data were synchronised through the Vicon system. The motion analysis system tracked 69 passive retro-reflective markers positioned according to a modified 6 Degrees of freedom (6 DoF) whole body model developed by Capozzo et al. (24). The kinematic model allowed for segmental kinematics to be tracked in 6 DoF (Figure 1).



Head markers:
- elastic head band with 4 markers

HEAD_BR	*..back right*
HEAD_FR	*..front right*
HEAD_BL	*..back left*
HEAD_FL	*..front left*

- 1 marker for static calibration only

HEAD_TOP	*..head top*

Trunk markers:

TOR_C7	*..7th cervical vertebra*
TOR_T10	*..10th thoracic vertebra*
TOR_CLAV	*..jugular notch*
TOR_STER	*..xiphoid process*
SCAPI	*..left scapula*
TOR_SHR/L	*..right/ left acromion*

Upper limb markers:

R/LARM_ELL	*..right/ left lateral elbow*
R/LARM_UP	*..right/ left middle upper arm*
R/LARM_ELM	*..right/ left medial elbow*
R/LARM_WRU	*..right/ left ulnar wrist*
R/LARM_WRR	*..right/ left radial wrist*
R/LARM_FIN	*..right/ left head of 2nd metacarpal bone*

Pelvic markers:

PEL_PSISL	*..left posterior superior iliac spine*
PEL_ASISL	*..left anterior superior iliac spine*
PEL_PSISR	*..right posterior superior iliac spine*
PEL_ASISR	*..right anterior superior iliac spine*
PEL_SAC	*..sacrum*
PEL_ILCL	*..left iliac crista*
PEL_ILCR	*..right iliac crista*
PEL_TROCHL	*…left greater trochanter*
PEL_TROCHR	*...right greater trochanter*

Lower limb markers:

R/LKNE_L	*..right/ left lateral femoral epicondyle*
R/LKNE_M	*..right/ left medial femoral epicondyle*
R/LTH_1, 2, 3, 4	*..right/ left 4 thigh markers on a cluster*
R/LANK_L	*..right/ left lateral malleolus*
R/LANK_M	*..right/ left medial malleolus*
R/LSHA_1, 2, 3, 4	*..right/ left 4 shin markers on a cluster*
R/LFOOT_H	*..right/ left heal*
R/LFOOT_1H	*..right/ left head of the 1st metatarsal bone*
R/LFOOT_5H	*..right/ left head of the 5th metatarsal bone*
R/LFOOT_1B	*..right/ left base of the 1st metatarsal bone*
R/LFOOT_5B	*..right/ left base of the 5th metatarsal bone*
R/LFOOT_TOE	*..right/ left tip of 2nd toe*

Figure 1. The static calibration image as captured in Vicon with the list of all the markers used. Markers were placed on specific anatomical landmarks either directly on the skin or onto tight fitting elastic clothing. There were 49 individual markers placed, 16 markers attached on 4 marker clusters (with 4 markers per cluster) for lower extremities and 4 markers on an elastic head band. Palpation was used to identify bony landmarks and the same researcher placed the markers in order to minimize the inter-researcher variability. The 4 lower extremity clusters were fastened by an elasticated band/ wrap.

Protocol

Participants were asked to ascend and descend stairs in a step-over-step manner at a speed dictated by a metronome set at 90 beats per minute to standardize for the potentially confounding influence of marked differences in walking speed. This speed was selected as it has previously been shown to correspond closely with the self-selected speed in elderly people during stair negotiation (92 + 10 steps/min)[19].

There were two conditions tested in the knee OA group while participants ascended and descend the stairs: 1. without the use of the handrail (OA) and 2. with use of the handrail on the contralateral side to the affected knee (OA + H). The rationale for using the contralateral handrail to the affected knee was due to the beneficial effect in terms of loading on the knee with walking stick and handrail use on the contralateral side as described in the introduction. The controls (CTR) were examined negotiating stairs only without use of the handrail. None of the participants used the handrails except for the OA + H condition. Three trials per condition, i.e. three ascents and descents, were recorded for each participant.

Data analysis

Kinematic data were labelled in VICON Nexus (Vicon Nexus 1.8.2) before both kinematic and kinetic data were exported to Visual3D software (Visual3D Student Edition v4.96.9; C-Motion Inc., Rockville, MD, USA) for biomechanical modelling. All lower extremity segments were modelled as rigid bodies. Anatomical frames were defined by landmarks positioned at medial and lateral borders of the joint, and for each segment a right handed co-ordinate systems was defined. The CoP was measured from each of the force platforms and the overall CoP when one foot was on two separate force platforms (during the double stance phase) was calculated by a weighted average approach as previously described[25]. The whole-body CoM was determined using the CoM of the individual segments and their positions, where the segment CoM contributions and positions were determined using a regression algorithm based on data measured by Dempster (26). The separation of the CoM from the CoP was then calculated in both anteroposterior (A-P) and mediolateral (M-L) direction (sagittal and frontal plane). I have compared the maximal separation and its range in the M-L direction; and the maximal anterior and posterior separation and its range in the A-P direction. The velocity of the CoM was calculated as the differential of the CoM displacement in M-L and vertical directions. Pain during stair negotiation was assessed using a 0-100-mm VAS after each condition, separately for ascent and descent.

A mean across the three trials per participant was calculated for each variable during the entire period on steps two to five; in case of the single leg stance time and stride width this was calculated as a mean of six values (two stance phases per trial; three trials). Only the affected leg (or the more severely affected leg in case of bilateral disease) in the OA group for both conditions and standardised to the right leg in the CTR group were analyzed for the single leg stance time.

Statistical differences between the groups (OA and CTR) for all parameters were compared separately for stair ascent and descent, and were tested using analysis of covariance (ANCOVA) with gait speed as a covariate. I used walking speed as a covariate, since although I tried to standardise for this, there were some small but significant differences between the groups during stair descent. The gait velocity, stride width and pain between the OA and CTR groups were tested using an unpaired samples Student’s *t*-test. Additionally, differences for all parameters between 2 conditions within the OA group, no handrail (OA) vs. contralateral handrail (OA + H) condition, were tested using a paired samples Student's *t*-test.

5.4 Results

Participant characteristics

As presented in Table 1 both groups were similar with regard to age, height, weight and BMI. The radiographic characteristics of the OA group are shown in Table 1. Information about the remaining 30 % of the OA sample was documented by arthroscopy or MR imaging (data not shown).

Table 1. Participant Characteristics*.

	OA	CTR	*p*
	n= 30	n= 30	
Age, years	58.9 (7.7)	61.6 (11.7)	*0.290*
Height, m	1.7 (0.1)	1.7 (0.1)	*0.450*
Body mass, kg	77.0 (15.9)	74.0 (10.9)	*0.409*
BMI, kg/m^2	27.4 (3.8)	25.8 (3.0)	*0.094*
Sex	57 % F	47 % F	
The radiographic characteristics of the OA group (%):			
K-L grade 1/2/3 in the PF joint	3.3/20/46.7		
K-L grade 1/2/3 in the TF joint	6.7/20/43.3		
Medial JSN** grade 1/2	30/40		
Lateral JSN grade 0/1/2	13.3/50/6.7		
Patellar JSN grade 1/2/3	43.3/20/6.7		

* Data are presented as mean (SD) except for sex. The p-value corresponds to an independent t-test comparing the two groups.

** JSN = joint space narrowing

† Significant difference between groups ($p < 0.05$).

Centre of mass velocity

During stair ascent, the mean M-L CoM velocity did not differ between the OA and CTR groups. However, the mean M-L CoM velocity was significantly reduced with handrail use compared to the condition without the handrail in the OA group (Table 2).

Table 2. Stair ascent: OA participants (OA), OA participants with contralateral handrail (OA + HR) and controls (CTR)*.

	Mean (SD)			*p*	
	OA	OA + HR	CTR		
	n = 30		n = 30	*OA vs. OA + HR***	*OA vs. CTR****
CoM velocity (m/s):					
Medio-lateral: Mean	**0.076 (0.015)**	**0.062 (0.013)**	0.074 (0.012)	***0.000 †***	*0.671*
CoM-CoP separation (m):					
Medio-lateral: Max	0.114 (0.027)	0.109 (0.022)	0.115 (0.020)	*0.223*	*0.609*
Medio-lateral: Range	0.200 (0.049)	0.190 (0.037)	0.198 (0.041)	*0.119*	*0.892*
Anterio-posterior: Anterior Max	0.128 (0.024)	0.124 (0.020)	0.132 (0.017)	*0.256*	*0.820*
Anterio-posterior: Posterior Max	**0.181 (0.026)**	**0.192 (0.026)**	0.176 (0.028)	***0.008 †***	*0.744*
Anterio-posterior: Range	0.309 (0.033)	0.316 (0.022)	0.308 (0.027)	*0.165*	*0.673*
Other variables:					
Single leg stance time (s)	0.843 (0.095)	0.840 (0.102)	0.814 (0.055)	*0.748*	*0.692*
Gait speed (m/s)	0.50 (0.05)	0.50 (0.05)	0.51 (0.03)	*0.947*	*0.163*
Stride width (m)	**0.11 (0.02)**	**0.10 (0.02)**	0.10 (0.02)	***0.023 †***	*0.078*
VAS (mm)	**24 (24)**	23 (23)	**0 (0)**	*0.381*	***0.000 †***

* Data are presented as mean (SD). † Significant difference ($p < 0.05$).

** Paired t-test

*** ANCOVA for CoM velocity, CoM-CoP separation and stance time; unpaired t-test for gait speed, stride width and VAS

During stair descent, the mean M-L and vertical downward CoM velocity did not differ between the OA and CTR groups. However, like stair ascent the mean M-L CoM velocity was significantly reduced with handrail use compared to the condition without the handrail in the OA group. The mean vertical downward CoM velocity between the conditions within the OA group was not different. However, when the peak vertical downward CoM velocity was compared during stair descent, the OA group had a significantly greater peak vertical downward CoM velocity compared to controls; handrail use did not lead to any changes in peak vertical downward CoM velocity within the OA group (Table 3).

Table 3. Stair descent: OA participants (OA), OA participants with contralateral handrail (OA + HR) and controls (CTR)*.

	Mean (SD)			*p*	
	OA	OA + HR	CTR		
	n = 30		n = 30	*OA vs. OA + HR***	*OA vs. CTR****
CoM velocity (m/s):					
Medio-lateral: Mean	**0.087 (0.012)**	**0.075 (0.015)**	0.084 (0.016)	***0.000 †***	*0.481*
Vertical: Mean	0.267 (0.031)	0.269 (0.027)	0.283 (0.018)	*0.510*	*0.793*
Vertical: Peak	**0.634 (0.085)**	0.621 (0.080)	**0.604 (0.061)**	*0.105*	***0.028 †***
CoM-CoP separation (m):					
Medio-lateral: Max	0.132 (0.021)	0.127 (0.022)	0.131 (0.020)	*0.138*	*0.935*
Medio-lateral: Range	**0.236 (0.040)**	**0.226 (0.037)**	0.229 (0.037)	***0.032 †***	*0.490*
Anterio-posterior: Anterior Max	0.127 (0.016)	0.130 (0.024)	0.125 (0.021)	*0.263*	*0.723*
Anterio-posterior: Posterior Max	0.202 (0.025)	0.200 (0.022)	0.215 (0.023)	*0.524*	*0.073*
Anterio-posterior: Range	0.329 (0.024)	0.331 (0.022)	0.340 (0.023)	*0.658*	*0.113*
Other variables:					
Single leg stance time (s)	0.816 (0.104)	0.804 (0.083)	0.765 (0.046)	*0.213*	*0.861*
Gait speed (m/s)	**0.49 (0.06)**	0.49 (0.05)	**0.52 (0.03)**	*0.677*	***0.009 †***
Stride width (m)	**0.15 (0.02)**	0.14 (0.03)	**0.14 (0.03)**	*0.125*	***0.043 †***
VAS (mm)	**31 (29)**	29 (27)	**0 (0)**	*0.131*	***0.000 †***

* Data are presented as mean (SD). † Significant difference ($p < 0.05$).

** Paired t-test

*** ANCOVA for CoM velocity, CoM-CoP separation and stance time; unpaired t-test for gait speed, stride width and VAS

CoM-CoP separation

During stair ascent, there was no difference in the CoM-CoP separation between the OA and CTR groups in the M-L (maximum and range) or A-P (anterior maximum, posterior maximum and range) directions. Within the OA group, the maximum posterior separation was significantly greater with handrail use compared to without. None of the remaining parameters in the M-L or A-P direction were different between the two conditions within the OA group (Table 2).

During stair descent, there was no difference in the CoM-CoP separation either in the M-L or A-P direction between the OA and CTR groups. Within the OA group, handrail use significantly reduced the range of the M-L CoM-CoP separation. The remaining parameters in the M-L or A-P direction were not different between the two conditions (handrail use *vs.* no handrail use) within the OA group (Table 3).

Single leg stance time

During stair ascent, there was no difference in single leg stance time between the OA and CTR groups or between the two conditions (handrail use *vs.* no handrail use) within the OA group (Table 2). Similarly, during stair descent, there was no difference in single leg stance time between the OA and CTR groups or between the two conditions within the OA group (Table 3).

Gait speed, stride width and pain

No significant differences were found between the OA and CTR groups when comparing the gait speed and stride width during stair ascent. However, as expected, the OA group was in significantly more pain compared to the CTR group. When comparing the handrail use vs. no handrail condition within the OA group, the use of handrail significantly reduced the stride width compared to no handrail, no differences between the two conditions were found when comparing gait speed or pain (Table 2).

Not only was the OA group in significantly more pain compared to the CTR group during stair descent, but the OA group also descended the stairs with a significantly wider stride width and at a slower gait speed compared to the CTR group despite the efforts to standardize the speed. The handrail did not have any effect on the gait speed, stride width or pain compared to no handrail within the OA group (Table 3).

5.5 Discussion

In the present study I investigated whether alterations to knee joint mechanics associated with knee OA caused impairments to balance during stair negotiation, and whether the use of the contralateral handrail improved balance control in people with knee OA. Based on the findings from the literature it is known that compared to healthy controls, knee OA population has reduced balance control as evaluated during functional tests and posturography[14-16]. However, an understanding of the effects of knee OA on balance during stair negotiation is missing. It was decided to test for any balance issues in the OA group during stair negotiation by assessing the velocity of the CoM and the separation between the CoM and CoP as these parameters can also indicate balance isssues during activities such as stair negotiation[21,27]. From a mechanical perspective, a more anterior position of the CoM in relation to the CoP (and hence greater CoM-CoP separation) during stair descent would mean a greater challenge to maintain balance. Additionally, the CoM velocity provides an unique insight to the level of "control" of dynamic balance during stair negotiation[21], with lower CoM velocity indicating more control over the CoM and a higher CoM velocity indicating a reduced control over the CoM. An example of the CoM and CoP trajectories, and the separation between the CoM-CoP for an OA participant during stair ascent is shown below (Figure 2).

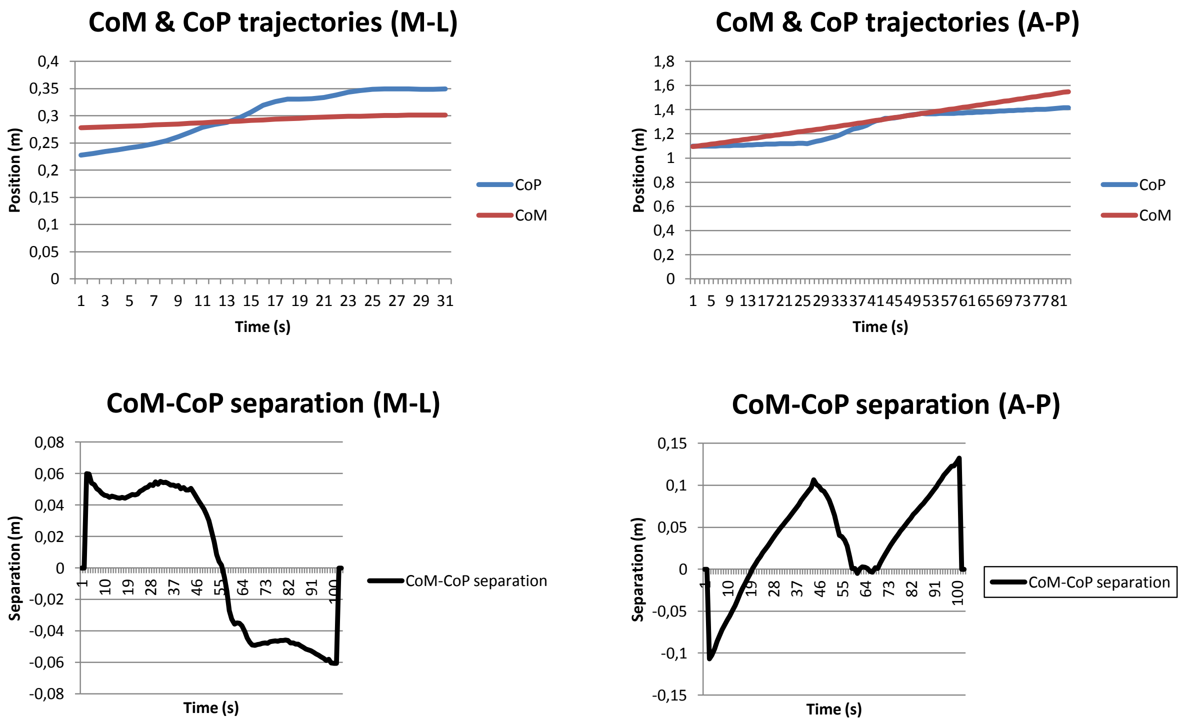


Figure 2. The CoM and CoP trajectories in mediolateral (M-L) and anteroposterior (A-P) direction, and the separation between the CoM-CoP in M-L and A-P direction in OA participants during stair ascent.

It was found that knee OA does not cause major balance impairments during stair negotiation as quantified by the CoM-CoP separation. Nevertheless, a higher peak vertical downward CoM velocity observed in the OA group during stair descent indicates that there is a specific phase of the gait cycle where control over balance may be compromised. This may be a limitation during the final part of the single limb lowering phase of the gait cycle and may pose a greater problem with increased stair riser heights and especially for more severely affected knee OA patients. Moreover, stair descent appeared to be a more challenging task for the OA group as they chose to descend stairs with a wider stride width and at a slower gait speed compared to controls. Also, within the OA group, a number of differences were identified during both stair ascent and descent with handrail use compared to no handrail. During stair ascent, handrail use caused a reduced mean M-L CoM velocity, greater CoM-CoP separation in the posterior direction, as well as a narrower stride width compared to no handrail. During stair descent, handrail use reduced the mean M-L CoM velocity and the range of the CoM-CoP separation in the M-L direction compared to no handrail. The pain experienced in the OA group during both stair ascent and descent was not alleviated by using the handrail. It is known that both static and dynamic standing balance are impaired in knee OA population based on poorer performance measured by posturography and functional tests (such as “Timed up and go”), which was interpreted as translating to poorer balance[15-17]. To my knowledge, this is the first biomechanical study to investigate balance control during stair negotiation in a mixed knee OA population compared to controls and to examine the influence of stair handrail use.

Centre of mass velocity

It was decided to look at the mean M-L CoM velocity during stair ascent and descent, and also the mean and peak vertical CoM velocity during stair descent to specifically focus on the way OA participants lowered themselves down onto the next step. It was hypothesised that a reduction in the CoM velocity might be a more cautious strategy to improve the stability of gait, as shown previously in a relatively weak (but non-OA) older population[21]. I expected that the OA group might negotiate the stairs with a greater CoM velocity as OA patients have previously been shown to display poorer balance during quiet standing[15,16]. However, I found that during both stair ascent and descent the mean M-L CoM velocity was not different between the OA and CTR groups indicating that the side-to-side movement of the CoM was not influenced by the pathology. This is an interesting finding as it might be expected that OA patients could prefer their healthy leg for weight-bearing and thereby potentially affect the M-L movement of the CoM. However, the peak vertical downward CoM velocity, which reaches its maximum towards the end of the single leg support phase just prior to the leading foot contacting the step below[21], was higher in the OA group, indicating more of a “free fall“ phase just before contacting the step below during stair descent. An explanation why OA patients seemed to find the final stages of controlled lowering down onto the next step challenging, likely relates to the stiffer and more painful knee joint, leading them to the situation when they could not sustain the required moment to execute the final part of the movement under control. I have previously observed a “stiffer gait“ employed by the same OA population during stair descent highlighted by less flexed knee joints when compared to controls[12] (Study 1/ Chapter 2). This may have implications for larger step heights, where the distance that the body needs to be lowered is increased compared to the standard situation in the present study. This may, therefore, lead to greater difficulties in the final phase of the gait cycle when lowering the body down onto the next step within similar or more severely affected patient groups.

The use of the contralateral handrail seemed to attenuate the mean M-L CoM velocity in the OA group during both stair ascent and descent confirming the hypothesis. Although this seems to be a useful strategy for minimising the velocity of the CoM in the M-L plane and hence making the task more stable, it may not be necessary in the OA group with moderate radiographic changes I have examined here (as they were similar to controls with regard to the mean M-L CoM velocity), but may become more important in a more severely affected OA population.

CoM-CoP separation

It was decided to assess the CoM-CoP separation in both the M-L and A-P direction by evaluating both the maxium separation and the range of the separation in both planes. The separation between the CoM and CoP provides a measure of dynamic balance during gait and the greater the CoM-CoP separation, the greater the demands on balance control[19]; a more lateral (or anterior) position of the CoM in relation to the CoP (i.e. greater CoM-CoP separation) leads to a more unstable body position in lateral (or anterior) direction. The CoM undergoes a cyclic lateral and vertical translation during stair negotiation as in level walking[20]. A recent study found that the medio-lateral CoM-CoP separation is larger during stair negotiation compared to level walking in healthy adults suggesting that stair negotiation is more challenging to balance than level walking[27]. The largest CoM-CoP separation representing positional instability occurs during single leg stance[20]. In a healthy elderly population, Reeves et al. (19) found a reduced peak CoM-CoP separation in the frontal plane in healthy elderly compared to the young during stair ascent. During stair descent, it was found that the A-P and M-L CoP separation were significantly lower in healthy elderly compared to young[28]. However, in constrast to my hypothesis, I found no differences in the CoM-CoP separation in the A-P or M-L direction between groups during stair negotiation, suggesting that knee OA does not lead to balance impairments quantifiable by the CoM-CoP separation.

The use of the handrail within knee OA patients showed some differences in CoM-CoP separation. There was a greater maximal posterior A-P CoM-CoP separation during stair ascent with the handrail condition compared to no handrail. This meant that the CoM was further behind the CoP with handrail use compared to no handrail, indicating that the OA participants were leaning back slightly more with the handrail. One possible explanation of why they chose this strategy is that relying on the handrail reduced their need to lean forwards to reduce the joint moment demands at the knee. During stair descent the M-L CoM-CoP range was reduced with use of the handrail compared to without confirming my hypothesis that the handrail would reduce the CoM-CoP separation and hence make the gait more stable in the lateral direction. Although the M-L movement of the CoM was reduced with handrail use, it is not expected to have major implications for balance control since this balance parameter was not compromised in the first instance in the OA group compared to controls.

Stance time, gait speed, stride width and pain

Although Hicks-Little et al. (18) found that patients with knee OA demonstrated less time in single support, I found no differences in the single leg stance times between the OA and CTR group or between the handrail and no handrail conditions within the OA group. This was surprising as it was expected that the OA group would try to limit the stance time on the affected (painful) leg to a minimum and hence differ from the CTR group and also due to the findings of impaired balance during single leg stance in the OA population[17]. The OA group was in more pain compared to controls, but the handrail did not alleviate the pain. Interestingly, despite their pain, the OA participants performed the task in almost the same manner as the healthy CTR group as the only differences between the two groups were a wider stride width, slower gait speed and a greater peak vertical downward CoM velocity in the OA group compared to controls during stair descent. Therefore having this level of pain does not seem to be a major factor in balance control during stair negotiation in this population. It must be noted that the present OA group was a relatively well-functioning sample as they were able to negotiate the stairs without handrail use during the unaided condition, which might not be representative of all OA patients.

The OA group descended the stairs at a slower speed and with a wider stride width as mentioned above similarly to the findings of Hicks-Little et al. (18). I hypothesise that the lower speed was chosen by the OA participants as the task was possibly more difficult for them than stair ascent. The wider stride width might have been employed to help them with balance, although I did not observe any differences between the groups when assessing the parameters chosen to evaluate balance. When the handrail was used during stair ascent, the stride width was narrower compared to no handrail condition in the OA group, possibly suggesting that the support through hand enabled the participants to narrow their base of support on the step and still remain stable.

Limitations and recommendations

Firstly, the mixed knee OA population in the present study might be a difficult patient group to draw conclusions from, which might be easier by having a group with unicompartmental disease. However, as stated previously, a mixed disease is often seen in clinics and therefore my study reflects clinical reality. Due to the fact that I used standardized walking speed in order to limit any potential confounding effect of gait speed on the chosen parameters, the results from the present study need to be interpreted with caution when referring to other findings with possibly different patient groups and settings. As mentioned previously my OA sample was a relatively well-functioning and not a severely disabled OA group. This aspect needs to be considered before applying my results to other OA population groups as it is known that patients with greater severity of knee OA may display more deficits in balance control compared to less severely affected patients[15,16].

With respect to future studies and recommendations, instrumented handrails would be useful to know how much force is being transmitted and how much the participants used the rails to help themselves during stair ascent and descent. As stair negotiation represents a considerable daily challenge for people with knee OA[9,29] and also a risk of falling in the elderly population[30], the evaluation of balance control and strategies aimed at preventing falls seems useful in knee OA patients[15,16]. Furthermore, future studies should establish a more direct link between the findings from the literature of a reduced balance control in the OA population based on clinical (functional) tests and/ or posturography with studies similar to the present one assessing the CoM and CoP, so that the relationship between the clinical findings and present results can be understood.

Conclusion

In conclusion, knee OA does not cause major balance impairments during stair negotiation, however, a higher peak vertical downward CoM velocity indicates that there is a specific phase of the gait cycle during stair descent where control over balance may be compromised in knee OA patients. This might indicate a poorer control of balance in the terminal phase of single leg stance phase during stair descent, with potentially more important implications for stairs with higher step heights and knee OA patients with greater severity. The contralateral handrail use in knee OA patients only minimally affected the CoM-CoP separation, although it did reduce the mean M-L CoM velocity.

5.6 References

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Chapter 6

Conclusions and future directions

6.1 Summary of main findings

In the present thesis I have evaluated lower limb biomechanics during stair negotiation in a mixed knee osteoarthritis (OA) population compared to healthy controls focusing on lower limb joint kinematics and kinetics, as well as parameters involved in balance control (Chapters 2 and 5). Furthermore, I have evaluated the effects of a patellofemoral (PF) knee brace and the use of handrails on knee joint biomechanics in the OA population (Chapters 3 and 4). These research questions were posed as a result of identifying the relevant gaps in literature as described in the thesis Introduction (Chapter 1). I believe that the findings from these four studies will contribute to scientific understanding in the field of knee OA and may translate through to clinical practice, where the patients who suffer from knee OA could benefit.

In my first study (Chapter 2) I investigated biomechanical characteristics in mixed knee OA participants with predominantly PF symptoms compared to healthy controls during stair negotiation. To my knowledge it was the first study to look at mixed knee OA compared to the more frequently investigated tibiofemoral (TF) OA and one of the few to look at “true” stair negotiation as opposed to a stepping task. Not surprisingly, the OA group experienced more knee pain than controls while negotiating stairs. However, major biomechanical differences were identified at the knee joint, as well as differences at the hip and ankle, which I believe were compensatory mechanisms enabling knee OA patients to cope with the altered knee biomechanics and pain. Lastly, differences between stair ascent and descent in both groups were identified. The key findings during stair ascent in the OA group compared to controls were a reduced peak internal knee extension moment, which might have been a response to the pain in the PF joint, and secondly the increased peak internal knee abduction moment, which is in line with the literature findings of an increased external knee adduction moment (EKAM) in the TFOA population during gait[1,2]. These results suggest that during stair ascent OA participants were able to reduce the PF load by decreasing the internal knee extension moment, but that was possibly done at the expense of an increased load on the medial knee compartment compared to controls. Similar findings during level walking and stair ascent were identified by Kaufman et al. (3) suggesting that it could be a strategy to minimize pain by reducing the internal knee extension moment and so minimizing the

PF loading. The key findings during stair descent in the OA group compared to controls was a reduced peak internal knee extension moment, which again might be reflecting a strategy to reduce the pain and load in the PF compartment in agreement with the findings by Kaufman et al. (3). Additionally, it may have been a way to put less demand on quadriceps muscle, which has been shown to be weaker in both early and established knee OA[1], or caused by an altered quadriceps force production during stair descent as suggested by Hinman et al. (4). The finding of a lower internal knee extension moment during stair ascent and descent was confirmed by a recent study that compared patients with PFOA, mixed disease of PF and TFOA and a control group during stair negotiation on a flight of 3 steps, with the internal knee extension moment found reduced in both knee OA groups compared to controls[5].

The second study (Chapter 3) investigated the effect of a PF brace on knee biomechanics within the mixed knee OA group (TF and PF compartments affected). It is known based on the literature that such a brace can reduce knee pain[6-8]. However, no study has previously evaluated its effect on knee joint kinetics and kinematics. The most important findings include a reduced peak internal knee extension moment during stair ascent with use of the brace, likely reducing the force required from the knee extensors and reducing the PF loading. The results confirmed the hypothesis that a PF brace use can lead to alterations in knee kinematics and reductions in the joint moments about the knee predominantly in the sagittal plane. It was speculated that the mechanism explaining these changes may be related to a greater perception of joint stability with use of the brace. This brace has previously been shown to cause reductions in PF pain when worn and assessed over a longer period[6,7], which may be at least partly attributed to the biomechanical changes shown here. Considering Study

1 results (Chapter 2) that showed how a knee OA group negotiated stairs with reduced sagittal knee joint moments compared to controls, the findings from Study 2 (Chapter 3) therefore suggest that this particular PF brace reduces the sagittal knee moment even further during stair ascent, reducing the PF load.

In my third study (Chapter 4) I evaluated the use of the stair handrail on the affected side and contralateral to the affected side compared to no handrail use within a knee OA group with mixed disease during stair negotiation. This was the first study to investigate these aspects on stairs and in this specific knee OA population. The study showed that handrail use can potentially lead to positive changes in knee joint loading by reducing the knee joint moments during stair negotiation in both sagittal and frontal planes. Therefore, the stair handrail proved to be effective both for the TF and PF components of the disease. The implications of handrail use for PFOA during stair ascent were that handrail use on either the affected or contralateral side reduced the sagittal plane knee joint moment compared to the unaided condition. This finding of a reduced sagittal plane moment with the use of handrails on either side is important as it is likely to reduce loading in the painful PF compartment and could therefore be a simple yet effective strategy to mitigate disease progression. Building on the findings from Study 1 (Chapter 2) where it was found that the OA group negotiated stairs with a reduced sagittal knee joint moment compared to controls, these results suggest that using the handrail on either side reduces the sagittal knee moment even further during stair ascent in the OA cohort, reducing the PF load even further. Interestingly, in a previous study, it was found that whilst bilateral handrail use reduced the ankle joint moment, it increased the peak knee extension moment during stair ascent compared to the unaided condition in thirteen healthy elderly (in contrast to the sample in the present study)[9]. The implications of handrail use for TFOA during stair descent suggest that handrail use on the contralateral side reduced the peak knee abduction moment compared to the unaided and handrail on the affected side conditions. Therefore, this strategy led to a reduction of the medial knee loading, presumably by aligning the body centre of mass (CoM) more directly over the stance leg. This adaptation would have reduced the lever arm of the ground reaction force in the frontal plane around the knee joint, to reduce the moment around the knee in the frontal plane. These findings are in line with the reduced frontal plane knee joint moments reported with contralateral walking stick use during level walking[10,11]. Therefore, contralateral handrail use on stairs, similarly to contralateral walking stick use on level ground, leads to unloading of the medial knee compartment. Contralateral handrail use on stairs could therefore be recommended as a strategy for patients with either medial TFOA or mixed disease with medial TF involvement such as the present sample to mitigate disease progression. In terms of the research on walking stick use, it was found that contralateral walking stick use led to a 10 % reduction of the EKAM in people with TFOA during level walking compared to no walking stick[10]. In comparison, on stairs I found a 15 % and 12 % reduction of the knee abduction moment by contralateral handrail use compared to the condition without handrail and handrail use on the affected side, respectively.

The fourth study (Chapter 5) of my thesis investigated whether there is an impairment in balance control in a mixed knee OA population during stair negotiation compared to healthy controls. Secondly, it assessed whether contralateral handrail use has any effect on balance control in the OA group. It was found that this particular OA sample had no major balance issues during this task when compared to controls. However, there were some differences between the two groups pointing to the fact that the OA group might have experienced difficulties especially during stair descent as indicated by a higher peak vertical downward CoM velocity, wider stride width and slower gait speed compared to controls. An explanation for why OA patients seemed to find the final stages of controlled lowering onto the next step challenging (indicated by the greater peak vertical downward CoM velocity), might relate to the stiffer and more painful knee joint, leading them to the situation when they could not sustain the required moment to execute the final part of the movement under control. I did observe a “stiffer gait“ employed by the OA population during stair descent highlighted by less flexed knee joints when compared to controls in Study 1 (Chapter 2). The use of the contralateral handrail reduced the mean mediolateral (M-L) CoM velocity in the OA group during both stair ascent and descent. Although this seems to be a useful strategy for minimising the velocity of the CoM in the M-L plane and possibly making the task safer, it may not be necessary in the OA group that was examined here as they were similar to controls with regards to the mean M-L CoM velocity. To my knowledge, this was the first study to look at balance control by investigating changes in biomechanical parameters during stair negotiation in a mixed knee OA population compared to controls and to examine the influence of contralateral handrail use during this task.

6.2 Further findings of interest

The other findings in Study 1 (Chapter 2) at the hip and ankle joints in the OA group compared to controls were possibly compensatory mechanisms to reduce the pain and movement at the knee reflecting the way the OA group negotiated stairs: generally in a “stiffer” manner compared to controls. A stiffer gait identified in the OA group compared to controls during stair ascent possibly due to pain was suggested to be a compensatory mechanism to minimize quadriceps loading and thereby reduce compressive forces at the knee[12]. Although I tried to standardize the speed, the OA group descended the stairs more slowly compared to controls potentially reflecting the level of difficulty they experienced with this task. The OA group also used a wider stride width compared to controls during stair descent. Stair descent is more challenging and demanding in terms of medio-lateral balance[13]. The wider stride resulted in greater knee moments in the frontal plane during descent compared to ascent as observed previously[14]. When stair descent was compared to ascent within the OA group, the peak internal knee abduction moment was 1.7 times higher; this difference was even greater in the control group with the moment being 2.8 times higher. This reflects the increased loading in the medial compartment during stair descent compared to ascent in both groups, as also shown previously[14].

In addition to the changes in the knee joint moments in Study 2 (Chapter 3) mentioned earlier, there were small yet significant differences in knee joint kinematics with the brace on compared to without the brace in both sagittal and frontal planes during stair negotiation. It was assumed that these kinematic changes could result from a greater perception of joint stability with use of the brace. In Study 3 (Chapter 4), the knee kinematics during stair ascent were affected by handrail use but to a lesser extent than the knee joint moments. Also, there was no difference in pain levels across the three conditions during either stair ascent or descent. It was hypothesised that it might have been due to the fact that the participants were already in some level of pain upon arrival and throughout the testing.

In Study 4 (Chapter 5) the use of the handrail within knee OA patients reduced the M-L separation between the CoM and centre of pressure (CoP) compared to no handrail. However, it is not not expected to have major implications for balance control since this balance parameter was not compromised in the first instance in the OA group compared to controls. The OA group was in more pain compared to controls and the handrail did not alleviate the pain. Interestingly, despite their pain the OA participants performed the task in a comparable manner to controls as the only differences between the two groups were a wider stride width and slower gait speed, similar to the findings of Hicks-Little et al. (15), and a greater peak vertical downward CoM velocity in the OA group compared to controls during stair descent. When the handrail was used during stair ascent, the stride width was narrower compared to the condition without handrail use in the OA group, possibly meaning that the handrail enabled the participants to narrow their base of support and still remain stable.

6.3 Limitations and recommendations for future studies

There were a number of limitations in the studies, which have already been mentioned in the individual chapters. Firstly, the mixed knee OA might represent clinical reality, but it may also make the interpretation of findings somewhat more challenging. Both TFOA and PFOA are two distinct conditions with different mechanisms involved[8]. Therefore, the present findings could be potentially replicated in either isolated TFOA or PFOA to see if these groups would differ from the present sample. Also, I believe that the present findings bring a mixed knee OA and PFOA to closer attention of the scientific community.

Secondly, the OA participants were quite a well-functioning cohort as they were all capable of abstaining from using the handrail during the conditions when they were asked not to use it (only in the case of instability). Therefore, the results could be replicated in more severely affected OA groups to see if they are in fact even more severely affected in the ways that I have observed. The present results should be interpreted with caution when referring to other findings with possibly different patient groups.

Thirdly, I did not examine or exclude participants with varus or valgus alignment, and it is known that malalignment plays an important role when it comes to joint loading measures, such as the external knee adduction moment[16,17]. On the other hand, these were participants with a mixed disease with predominant PFOA and not exclusively medial compartment TFOA, which is associated with varus malalignment. Similarly, I did not examine or exclude participants with patella malalignment, which is associated with PFOA progression[18].

Continuing on the note of clearer structural characteristics of the study population, despite the fact that the control participants did not experience any knee pain or problems with their lower limbs, radiographic evidence of no knee OA was missing in the control sample. It is known that OA symptoms do not necessarily correspond with radiography and vice versa[19,20] so it cannot be assumed with 100 % certainty that the control group did not have some degree of radiographic knee OA. Based on the data discussed in the Introduction (Chapter 1), a “worst case scenario” estimate is that around 20 % of my sample could be with no symptoms but actually have some evidence of radiographic OA according to the studies that found symptomatic knee OA to be present in around 10% of patients and radiographic knee OA in around 30% of patients[7,21-24].

It should be cautioned that the kinematics models associated with motion analysis systems may not be as sensitive for measuring the movement of the knee joint in the frontal plane, especially considering the role of soft tissue artefact. Since some of the results were small in magnitude, as is the case for previous studies, they should be interpreted with caution. Furthermore, as discussed in Study 1 (Chapter 2), the present model was a 6DoF full body model. Although it appeared superior to the Plug In Gait model when tracking the movement of the knee joint for our purposes (considering the brace in Study 2/ Chapter 3 and having it standardised across all studies), which was the main joint of interest, it still has issues related to the interpretation of our results. For example, the validity of such a model for using the TF moments to make conclusions about the TF and especially PT loading remains questionable. The loading would be better estimated by looking at joint contact forces. The PF loading in particular is an issue as I have used the model only to measure the sagittal and frontal kinematics and kinetics at the knee (i.e. TF joint axis), which might have overestimated our interpretation of PF loading. The golden standard as far as models are considered to represent the actual movement at the knee remains to be determined as the knee movement is complex. The important take-home message is to always note what model a particular study used when comparing the magnitudes of results and be aware of the limitations of individual models and approaches. Furthermore, experimental studies using instrumented implants indicate the need to evaluate the simultaneous changes in sagittal and frontal plane kinematics to have a valid estimate on the effect of TF loading[25].

Muscle activity could have been recorded using electromyography (EMG), and might have added something to my understanding of muscle activations around the knee.

However, it was very difficult and in most cases not possible, to fit the electrodes under the tight-fitting knee brace (Chapter 3), it was chosen not to record EMG across all four studies.

Regarding Study 3 (Chapter 4), it would have been of interest to measure forces exerted on the handrail. I could not measure these forces as the handrails were not instrumented and so I was unaware how much force was being transmitted and how much the participants used the handrails to help themselves during stair ascent and descent. Although forces could not be measured directly on the handrails, the effects of handrail use would ultimately unload the lower limbs and I could therefore make fairly accurate inferences based on changes in lower limb joint moments.

In general there is a debate as to how to deal with the situation of two groups of people who likely walk at different speeds. The choice of using standardised walking speed in order to limit any potential confounding effect of gait speed on the chosen parameters, which was shown to be the case previously in the literature[26], was under scrutiny during the revision process when submitting the respective studies for publication. I captured a self-selected speed condition both in the OA and control group during stair negotiation, but since it showed exactly the same results as presented in this thesis for the standardized speed, the standardized speed was chosen as the better option of the two since it was controlling for this potentially confounding variable of gait speed. In fact, looking at the research done in the field of knee OA, researchers use both standardized and selected speeds. It seems to be a matter of choice and will likely continue to be an area for debate. Therefore when comparing study results, the specific experimental set up should always be considered when drawing conclusions.

With respect to future studies and recommendations gained from my experience on this thesis, I would suggest the following, considering the above-mentioned limitations. It may seem beneficial to have a clearly defined study population, for example a uni-compartmental disease over a mixed knee OA disease. However, on the other hand, it seems that a mixed knee OA group is within the broad OA patients´ spectrum that did not get much attention to date, so it should possibly receive further investigation. Frequently in the literature, however, studies looking at TFOA do not specifically state whether the PF compartment was involved or not, so it is perfectly possible that the participants in previous studies actually do have a mixed knee OA disease and that it simply was not clearly stated in those papers.

Lastly, future studies could evaluate the effect of insoles and other types of braces. There is a large variety of such devices[27] and and there has been very little investigation of their efficacy during stair negotiation. Also, compensatory mechanisms and gait retraining techniques such as trunk lean could also be investigated on the stairs to see the effect on knee joint biomechanics and loading[28,29]. Exercise and weight loss studies should be implemented when trying to replicate present study results to see their potential positive effect on the task performance, pain and lower limb joint biomechanics.

6.4 Conclusions and clinical implications

The importance of my thesis lies in underlining the fact that knee OA is a broad diagnosis that should be specified to a compartment and more studies should focus on mixed PF and TFOA or isolated PFOA as these patients represent different clinical realities from already quite extensively documented TFOA. The focus of future studies should be on patients with different compartmental involvement and on finding new mechanisms to reduce higher knee joint loading when present. These new strategies and mechanisms should then be applied in disease management in a subject-specific and/or compartment-specific way.

A highlight of this thesis is the set up with the seven-step staircase playing a central role, simulating natural stair negotiation as much as possible in a laboratory setting. Not many studies to date have access to a proper staircase and instead evaluate stair negotiation using a stepping task[5]. Stair negotiation represents a considerable daily challenge for people with knee OA[14,30] and a risk of falling in the elderly population[31], and so the evaluation of strategies aimed at preventing falls on stairs seems useful in knee OA patients[32,33]. Therefore the task of stair negotiation should be imitated by researchers as closely to daily reality for these patients as possible.

The novelty of Study 2 (Chapter 3) with the brace lies in the combination of the study group, experimental set up (both mentioned above) and the type of the brace. There are no studies that have investigated biomechanical effects of a PF brace as opposed to purely symptomatic pain-relieving effects[6,7] and no studies that have tested this type of brace on stairs.

In study 3 (Chapter 4), the idea to use a handrail during stair negotiation as a means of support originated from research on healthy individuals and in a knee OA population during level walking with the use of a walking stick. However, translating the positive effect of handrail use identified by this study to daily life might be problematic since handrails might not always be present on either side when negotiating stairs. Furthermore, it does not need to be only stairs that present a challenge for knee OA patients, but it could easily be ramps or pavements with inclines that these patients have to deal with during the activities of daily living. Inclined and declined ramps represent a different biomechanical challenge to stair negotiation. Therefore, future research should focus on strategies of how to make these activities easier and less painful for knee OA patients.

Findings from Study 4 (Chapter 5) revealed that the final stages of knee OA patients lowering themselves down onto the next step was problematic. This may have implications for larger step heights, where the distance that the body needs to be lowered is increased compared to the standard situation here, and may therefore lead to more difficulties in the final phase of the gait cycle when lowering the body down onto the next step. This problem might also be more of an issue for more severely affected knee OA patients.

As mentioned previously, the end-stage treatment for knee OA is total knee replacement[8]. Therefore efforts should be directed towards finding simple, accessible, effective and cost-effective conservative treatment strategies to prevent patients reaching this level. The various biomechanical interventions that are available for this patient group together with exercise, weight loss and education should remain at the centre of all therapeutic interventions in a knee OA population[34].

In conclusion, it was found that a mixed knee OA population with predominantly PF symptoms differs in lower limb biomechanical parameters when compared to healthy controls during stair negotiation at all three lower limb joints, but with major differences observed at the knee. Differences between stair ascent and descent in both groups were also identified. The knee OA sample did not have major balance impairments during stair negotiation compared to controls, however, stair descent is a more challenging task than stair ascent and some indications of a poorer balance control during the terminal phase of single leg stance were present. The contralateral handrail use in knee OA patients only minimally affected the balance control during stair negotiation. The PF knee brace led to small yet significant changes in knee joint angles and moments, mainly in the sagittal plane, but also to a lesser extent in the frontal plane during stair ascent and descent, potentially having a positive effect on knee loading. Handrail use proved to be an effective strategy for favourably modifying knee joint loads during stair negotiation for both the medial TFOA and PFOA aspect of the mixed knee OA sample.

6.5 References

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